

Chapter 2

A New Method for the Quality Assurance of Strength Measurements

Abstract This chapter describes the methods and the results of a research project aimed at the development of a modern protocol for the quality assurance of clinical strength measurements conducted by the handheld dynamometer (HHD). As several reliability issues in HHD measurements were raised in the literature, such analysis is needed in order to assess the quality and reliability issues occurring when measuring the maximum voluntary forces and moments exerted by human muscles. The chapter begins with a literature review about the methods commonly adopted to measure strength in clinical contexts and some modern works that involved the use of the HHD. Then, the methods of the novel protocol are described. The designed protocol takes advantage of an optoelectronic system to measure the HHD positioning with respect to the patient, its undesired motion during the trial, and an overall index of trial quality. The preliminary design and setup of the protocol is presented. Preliminary results are discussed as well as the limitations and issues encountered in the first design. Then, the final setup is presented, as well as the results of two campaigns of measurements conducted on knee and ankle strength assessment on adult healthy subjects. In knee strength measurements, the most relevant source of inaccuracy is identified in the angular displacement on the horizontal plane and the use of a single-component HHD induces an overall inaccuracy of $\sim 5\%$. Knee extension trials are the most critical due to the higher force exerted. In ankle strength assessment, the most relevant source of inaccuracy is the angular displacement on both the sagittal and horizontal planes and the worst results are observed for plantarflexion trials. Thus, the HHD measurement method is not recommended for evaluating ankle plantarflexion strength. The proposed protocol may be used in clinical contexts for the quality assurance of HHD strength measurements and in those cases where high accuracy of measurements is essential.

Keywords Ankle strength • Handheld dynamometer • Knee strength
Load cell • Lower limb • Manual muscle testing • Quality assurance

2.1 Introduction

Measuring strength means measuring the maximum voluntary contraction force produced by muscles.

Strength measurements are popular in medical practice as they provide information about the health of muscles and ligaments and document the effectiveness of training and rehabilitation programs (Allen et al. 1995; Bohannon 1990; Hughes et al. 2001; Maughan et al. 1983). Indeed, the joint force and torque estimation inherently describe the stability and health of the joint itself (Brunner and Rutz 2013).

The study of forces and torques is also used to assess the effects of neuromotor or genetic diseases on the musculoskeletal system. Examples are cerebral palsy, Prader–Willi syndrome, and Ehlers–Danlos syndrome, that are characterised by gait and muscular disorders, due to poor joint stability and muscle–tendon weakness (Ancillao et al. 2012; Brunner and Rutz 2013; Galli et al. 2011). Moreover, it was observed that obesity may have effects on the muscle power of lower limbs, influencing everyday tasks such as rising from a chair or walking (Capodaglio et al. 2009). It is clear that strength assessment plays an important role for the study of the previously cited pathologies and definition of rehabilitative treatments.

Several methods to measure human strength were developed across the years. The simplest ones were based on the indirect measurement of muscle force and fatigue as in the chair-stand test (Csuka and McCarty 1985). The procedure consisted in measuring the time required to stand up and sit back on a chair when the trial was repeated 10 times. As direct methods are, in general, to be preferred (Allen et al. 1995; Jones et al. 1999) some complex devices were designed over the years (Amundsen et al. 1987; Peindl et al. 1997). Such devices measured strength by using ropes, cantilevers, and the like which deviate the subject's force to some ad hoc designed force sensors. For instance, such systems were used to measure the strength of the knee extensor (Maughan et al. 1983; Fig. 2.1) and of the triceps brachii (Allen et al. 1995; Fig. 2.1) (Fig. 2.2).

Nowadays, the gold-standard method to measure strength is the isokinetic dynamometer, which is widespread, commercially available, and allows the estimation of the maximum force exerted by the patient during a specified exercise (Janssen and Le-Ngoc 2009; Kim et al. 2014; Martin et al. 2006; Tsaopoulos et al. 2011). The isokinetic dynamometer (Fig. 2.3) is composed of a seat and a moving arm which is instrumented in such a way as to gather the direct measurement of the forces and torques applied by the subjects. The resistance exerted by the lever arm can be dynamically adjusted, allowing the measurement of the muscular forces in dynamic conditions and providing optimal loading of the muscles (Baltzopoulos and Brodie 1989). Isokinetic dynamometry can also be used for the training of various muscle groups in order to improve muscular performance in dynamic conditions, as the motion of different activities can be simulated to improve the training effect (Baltzopoulos and Brodie 1989).

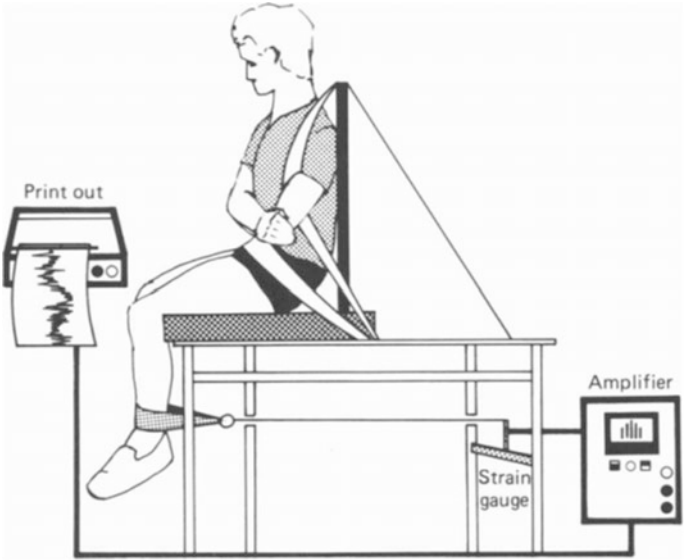


Fig. 2.1 Knee extension strength measurement (Maughan et al. 1983)

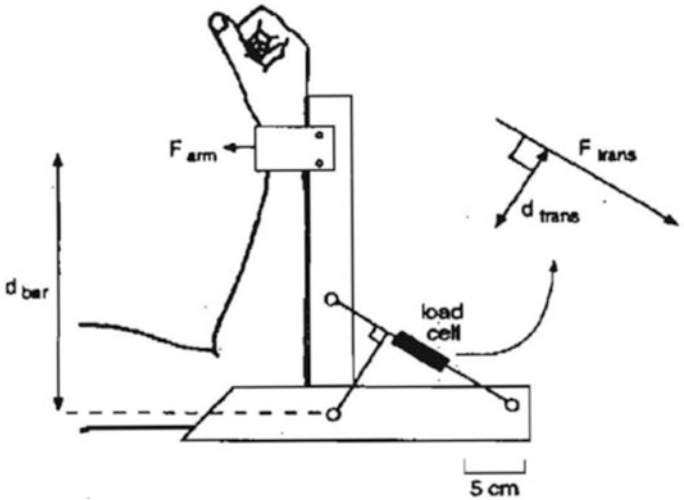


Fig. 2.2 Triceps brachii strength measurement (Allen et al. 1995)

The isokinetic dynamometer showed high interrater and intrarater reliability and reproducibility for the measurement of joint forces and torques when it was applied to subjects with different ages, both on the lower limb and upper limb (Fulcher et al. 2010; Hartmann et al. 2009; Hughes et al. 2001; Kim et al. 2014).



Fig. 2.3 Isokinetic dynamometer (www.biodex.com)

The main drawbacks of the isokinetic dynamometer are that it is expensive, it is not portable, and it requires a long time to prepare the subject as well as dedicated spaces. Moreover, during movements in the vertical plane, the torque registered by the isokinetic dynamometer is affected by gravitational forces and may contain artefacts due to inertial forces. These artefacts can be corrected by computer systems that allow the accurate computation of isokinetic parameters and real-time display of the torque output (Baltzopoulos and Brodie 1989).

An alternative and modern method to measure strength directly is by using a handheld dynamometer (HHD). It consists in the use of a small portable dynamometer that can be held in the hand by a clinician (Fig. 2.4) who applies it on some defined landmarks while asking the patient to exert a force against the dynamometer (Fulcher et al. 2010; Kim et al. 2014).

The HHD is a low-cost device, if compared to the isokinetic dynamometer, portable, easy to use, and does not require long-lasting procedures or dedicated rooms for its use. Moreover, the HHD allows the indirect measurement of joint torque by knowing the distance between the positioning landmark and the anatomical joint. Maximum joint torque can then be obtained by multiplying the maximum force measured by the distance from the joint centre. The HHD may be designed with different shapes, depending on the anatomical landmark to which it is intended to be applied and may exploit different working principles. Examples are: (i) spring/cantilever systems (Fig. 2.5a); (ii) hydraulic systems (Fig. 2.5b, c), and (iii) piezoelectric or strain gauge load cells (Fig. 2.5d). Most modern commercial HHDs are based on single-component load cells (Fig. 2.5d) that are relatively inexpensive and allow a fast measurement, data recording, and storage. However,

Fig. 2.4 Handheld dynamometer in use



correct application by the operator is crucial to ensure that the force is effectively directed on the sensible axis of the load cell and the subject's limb remains still during the measurement.

Two methodologies to measure strength by means of the HHD were described in the literature (Bohannon 1988): (i) the 'make' test, in which the examiner holds the dynamometer stable while the subject exerts a maximal force against it, and (ii) the 'break' test, in which the examiner holds the HHD in place but she or he has to overcome the maximum force exerted by the subject, consequently making the limb move in the opposite direction. The two methods were compared by Bohannon (1988) concluding that both were reliable and repeatable if the examiner had enough force to counteract the force exerted by the patient. The two methods were also comparatively examined by Phillips et al. (2000). The main outcomes of that work were: (i) the 'break' test requires a larger force exerted by the examiner, therefore weaker examiners may experience trouble and when this occurs the 'make' test is preferable; (ii) both 'break' and 'make' tests provide similar results, both are strongly operator-dependent, and, as a tendency, they underestimate the maximum force exerted by the knee extensor.

The reliability and repeatability of the HHD have been widely studied in recent years (Clark et al. 2010; Kim et al. 2014; Martin et al. 2006; Wuang et al. 2013). Wuang et al. (2013) focused the study on the measurement of strength of lower limb muscles in children with intellectual disabilities. The authors concluded that the use of HHDs could be considered practical and easy for clinicians but the measurement protocol had some critical issues related both to the operator's training and to the positioning of the dynamometer on the subject's limb. The operator's influence on the strength measurement was tested by Kim et al. (2014) by comparing three set-ups: (i) with the HHD fixed to the distal tibia by a Velcro

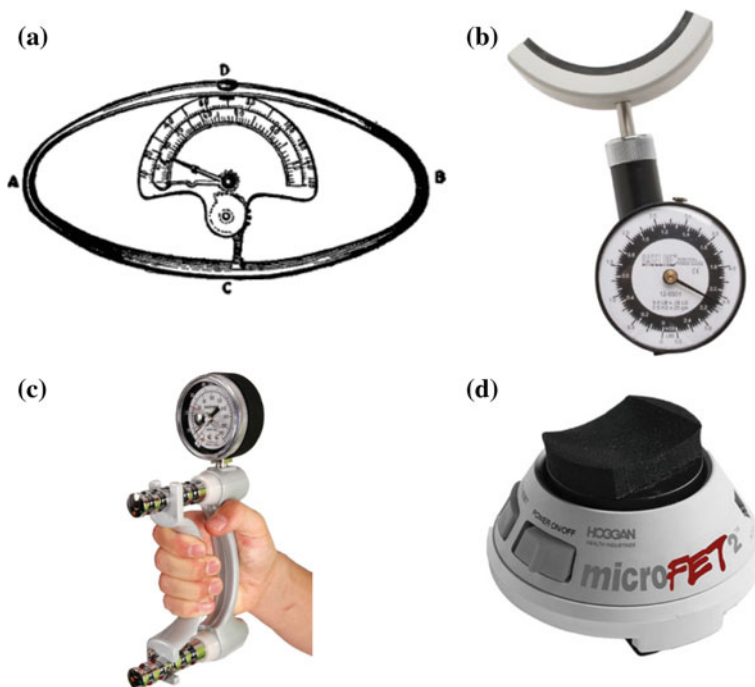


Fig. 2.5 Different designs of HHDs according to their scope. (a) Handgrip strength, based on a circular elastic element (ABCD). The force is applied between the points C and D; (b) general purpose, based on hydraulic system; (c) handgrip strength, based on hydraulic system; (d) general purpose, based on electronic-load cell system (Wagoner 1992)

strap, (ii) with the HHD held by the operator, and (iii) with an isokinetic dynamometer, assumed as a reference. They found that fixed and nonfixed methods showed good interrater reliability and the reliability of the fixed method was the highest.

In spite of the advantages of the HHD, reports on reproducibility and interoperator repeatability were controversial (Bohannon and Andrews 1987; Hébert et al. 2011; Marmon et al. 2013; Riddle et al. 1989). The main causes of low reliability of the HHD method were identified in the poor training of operators and incorrect patient positioning (Bandinelli et al. 1999). In fact, the HHD method relies on the operator's strength and training in order to contrast the force exerted by the patient (Bohannon 1988). According to Martin et al. (2006), the HHD offers a feasible, inexpensive, and portable method to test for quadriceps muscle strength in elderly people, but it was proved to underestimate the absolute quadriceps force if compared to the isokinetic dynamometer, especially when the subjects were stronger.

Whereas strength can be assessed for all human muscles, a particular clinical relevance is conferred to lower limb muscles, due to the important role they play in day-living tasks (walking, chair rise, climbing, etc.), which may be compromised by neuromotor pathologies and aging (Hartmann et al. 2009). Among the lower

limb joints, special attention should be paid to the ankle, as plantar flexion/extension and inversion/eversion are important determinants of balance and general functional ability (Spink et al. 2011). Moment exerted by the ankle, as well as ankle power and stiffness play an important role in human gait. In fact, ankle kinematics and kinetics are commonly affected by motor pathologies and may improve in the case of therapy (Camerota et al. 2015; Galli et al. 2008; Rigoldi et al. 2012; Vismara et al. 2016).

Several studies were conducted to assess reliability of ankle strength measurements based on the HHD method. For example, ankle strength of healthy subjects was measured by means of the HHD (Fig. 2.6) and then compared to an electromechanical dynamometer, that is, a fixed dynamometer that allowed evaluation of isometric force (Marmon et al. 2013). That study showed that HHD measurements were not correlated to the fixed dynamometer, assumed as reference, and statistical differences were found between the two sets of measurements. This was attributed to the low strength of the examiner and his inability to position and hold the HHD steady. The conclusion was that HHD strength measurements of the plantar-flexors should not be considered valid (Marmon et al. 2013). Such results were in disagreement with the results observed by Spink et al. (2011), who found high reliability of ankle and foot HHD strength measurements in older and younger participants. They concluded that the HHD is a valid methodology for the evaluation of ankle strength. Another work about HHD reliability on ankle measurements is the one by Hébert et al. (2011). They found that among all the lower limb joints, ankle plantarflexion and ankle dorsiflexion had the lowest reliability, therefore they recommended further study in this direction, especially regarding strength evaluation in children with neuromotor disabilities.

From the previously cited studies, the operator's inefficiency in holding the HHD in the right position emerged as the main issue in HHD strength measurements related to the ankle joint. In all the reported studies only a reliability analysis was conducted and, to the best of the author's knowledge, no studies were performed to identify and quantify the sources of inaccuracy that occur in the assessment of ankle strength by means of a HHD. Therefore detailed studies about the quality of clinical measurements are strongly encouraged (Mokkink et al. 2010)

Fig. 2.6 Position for the ankle plantarflexion strength testing by means of the HHD. Adapted from Marmon et al. (2013)



with the purpose of establishing reliability, reproducibility, and validity of such measurements (Terwee et al. 2007).

What emerges from the literature survey is that the operator's skills in holding the HHD in place represent a decisive factor on the quality of the strength measurements. Moreover, reliability studies conducted on the HHD produced controversial results as sometimes it was found to be 'excellent' (Mahony et al. 2009) and sometimes it was found to be 'low' (Verschuren et al. 2008).

A further unaddressed potential limitation in the use of HHD is that commercial HHDs acquire only the component of the force projected on the sensitive axis of the HHD and, consequently, the lateral components of force and the moments exerted by the subject on the HHD are always ignored.

Previous reliability studies were limited to the statistical analysis of repeated measurement and, to the author's best knowledge, no other studies were conducted to quantify the effect of the two previously cited limitations and directly measure the sources of inaccuracy occurring when commercial HHDs are used.

In the present work, the author describes the design and implementation of a modern motion capture protocol for the quality assurance of clinical strength measurements and for the direct measurement of the sources of inaccuracy that occur when strength is assessed by HHD targeting the knee (Ancillao et al. 2017b) and ankle (Ancillao et al. 2017a) joints. The method was validated on healthy adult subjects.

The following quantities are measured:

- Actual dynamometer position with respect to the joint
- Undesired motion of the limb
- Actual forces and torques exerted between patient and operator

2.2 Preliminary Setup

As a first step of the work, some preliminary testing and data acquisition were conducted in order to identify the best protocol design and the major issues occurring in data acquisition (Ancillao et al. 2015). The preliminary analysis was conducted in the Motion Analysis and Robotics Laboratory of 'Bambino Gesù' Children Hospital of Rome, Italy.

Subjects were recruited according to the inclusion criteria: healthy adult subjects of both sexes from 18 to 35 years old. Subjects must not have any neurological or orthopaedic disorders and must not have undergone surgery to the lower limb joints. All the subjects were evaluated by a trained clinician before inclusion.

Ten healthy adult subjects were enrolled in this study: 6 males, 4 females, mean age 27.3 ± 1.4 years, mean height 169.2 ± 11.2 cm, average body mass 65.4 ± 11.2 kg. They were all right-handed even though this was not an inclusion criterion.

To track position and motion of the subject, the operator, and the force sensor, we used a Vicon[®] MX Optoelectronic System (Oxford Metrics, UK) equipped with 8 IR cameras and strobes, Nexus 1.7 software, sampling frequency of 200 Hz, and a calibrated volume of about 4 m³. The overall inaccuracy was estimated as ~ 1 mm.

Force was measured by means of a commercial MicroFet[™] handheld dynamometer (Fig. 2.7, Hoggan Scientific, Salt Lake City, US). The dynamometer was set up to transfer real-time data by means of a built-in wireless connection. HHD had a sampling frequency of 100 Hz, maximum load capacity of 1300 N, and accuracy 1% of full range.

According to classical clinical protocols, the HHD is meant to be held in-hand by a trained clinician who applies it on some defined landmarks on the subject's limb and asks the patient to exert a force against it. The result of the measurement is the maximum force that occurs during the trial and the effective duration of the trial, which should be about 5 s (Bohannon 1986; Fulcher et al. 2010; Kim et al. 2014; Martin et al. 2006; Phillips et al. 2000).

The working principle of this HHD is based on a single-axis load cell that collects and records the force applied over the sensing surface (Wagoner 1992).

A microFET2(r) HHD was equipped with reflective markers, whose diameter was 10 mm, in order to reconstruct its position and orientation within the calibrated volume of the Vicon System. Three markers were placed on sticks and a fourth marker was placed at the centre of the sensing surface, as shown in Fig. 2.7. The central marker was necessary to reconstruct a reference position for force origin on the sensing surface. It was used only for the static trial and it was removed during the force measuring trial. The reconstruction was made possible by a localisation procedure based on a local reference system built on the three fixed markers (Ancillao et al. 2013).

The subject was equipped with a marker set based on the Plug-in-Gait (PIG) protocol with the addition of markers on the internal epicondyle of the femur

Fig. 2.7 The microFET2(r) handheld dynamometer equipped with passive motion capture markers



and of the ankle, in order to locate the knee and ankle joint centres easily and a cluster composed of three markers applied on the thigh (in correspondence with the quadriceps femoris) that allowed optimal reconstruction of thigh markers in case they were covered.

Markers were applied on both sides of the body even though strength was measured only on the dominant side. The full marker protocol, including markers on the HHD, is depicted in Fig. 2.8.

The markers on the HHD were placed on rigidly fixed sticks in order to keep the markers away from the HHD body and avoid covering by the operator’s hand (Fig. 2.7).

The designed protocol required the acquisition of a static trial before the set of dynamic trials. In the static trial the subject was asked to stand up motionless in the calibrated volume for about 5 s. The HHD was also kept still in the calibrated volume (Fig. 2.9). This allowed the recording of the position of the central marker, referencing the sensing surface, with respect to the other three markers. After calibration, measurement trials were recorded. The maximum voluntary force was measured by the HHD equipped with markers held and placed (Fig. 2.10) according to a standard clinical protocol, described in the following.

For an accurate analysis, the force signal generated by the HHD needed to be synchronised with the kinematic recording of motion. A first attempt to collect real-time data produced by the HHD and to synchronise it with kinematic data was

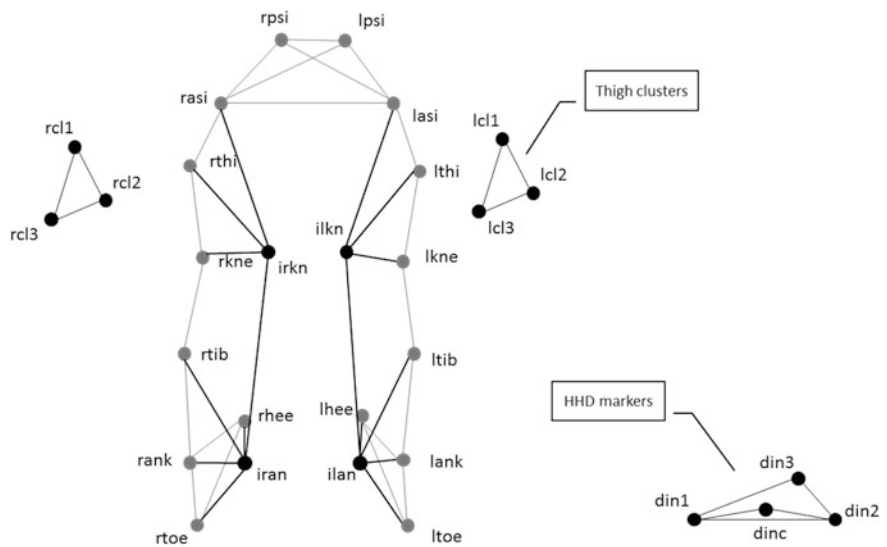


Fig. 2.8 The full marker set used for preliminary trials. Gray lines/dots represent the PIG model; black lines/dots are the markers added to the PIG

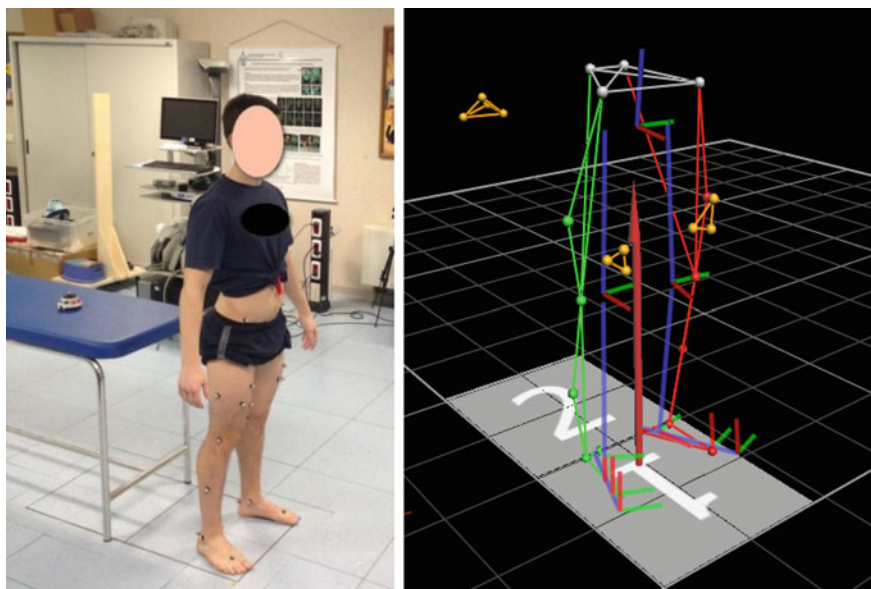


Fig. 2.9 Static trial for the preliminary protocol and its biomechanical reconstruction. The subject is wearing the full marker set

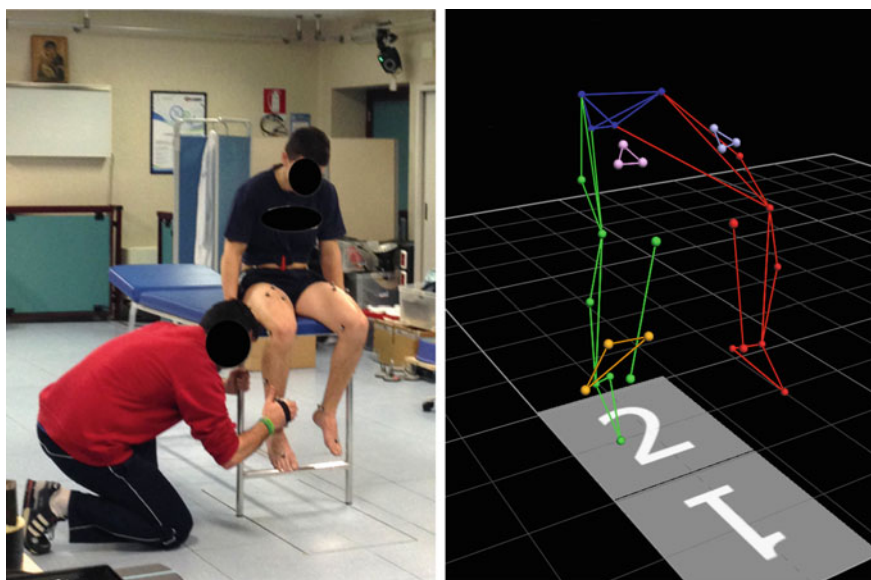


Fig. 2.10 Knee extension strength measurement trial for the preliminary protocol and its biomechanical reconstruction

done. As the HHD transmits data through a USB dongle, its driver was implemented in a LabView[®] programming environment (National Instruments, USA). Data collected by the dongle was streamed towards the analog output port of a National Instruments[®] A/D board (Fig. 2.11). The board was opportunely programmed to work as a digital-to-analog converter. Therefore an analog signal, carrying the force information recorded by the HHD, was generated in real-time and sent to the analog input port of the Vicon system through a wired connection. Testing on synchronisation performance was done by simultaneously applying a variable force in series to the HHD and the force platform connected to the Vicon system. This was achieved by vertically pushing the HHD against the surface of the force platform. This testing highlighted several critical issues in this synchronisation method. The most relevant issue was a stochastic delay in the force signal (Fig. 2.12) and a strong noise superimposed on the HHD signal itself. The delay was attributed to the computing time required by the PC to collect and process data and to the conversion speed/buffering limits of the A/D board. The delay was also probably due to delays in the commercial closed source wireless protocol implemented by the manufacturer of the HHD. As it was not possible to correct or predict the delay (and after some testing), the idea of automatic data synchronisation of the HHD data stream with the kinematic data through the analog port was dropped.

A procedure for manual synchronisation was implemented instead: kinematic signals and HHD signal were recorded separately. The HHD signal was collected by means of a LabView ad hoc designed software that saved the HHD data stream data to a CSV (text) file. Timestamps of packets were saved to ensure time consistency. Data files recorded by the two systems were imported in a MatLab(r) environment and were manually synchronised by the user who had to identify events visually on both signals. The events to be identified were the beginning (rising) and ending (descending) of the force track and the minimum distance between the HHD and the limb. This ensured an acceptable synchronisation between kinematic recording and force measurement.

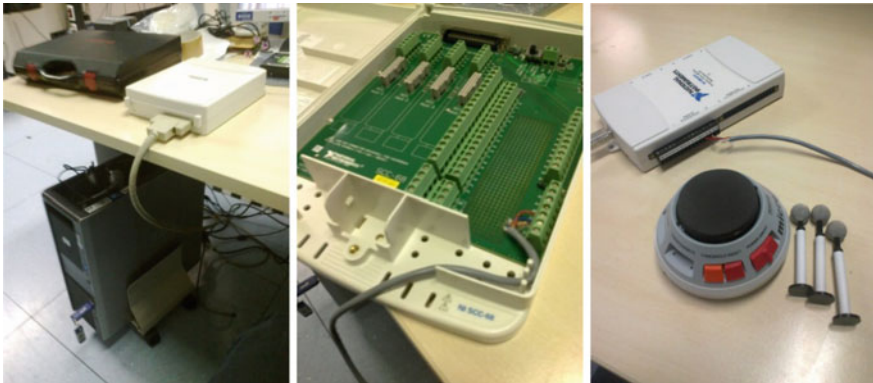


Fig. 2.11 System for collecting wireless HHD data and streaming to Vicon as an analog signal

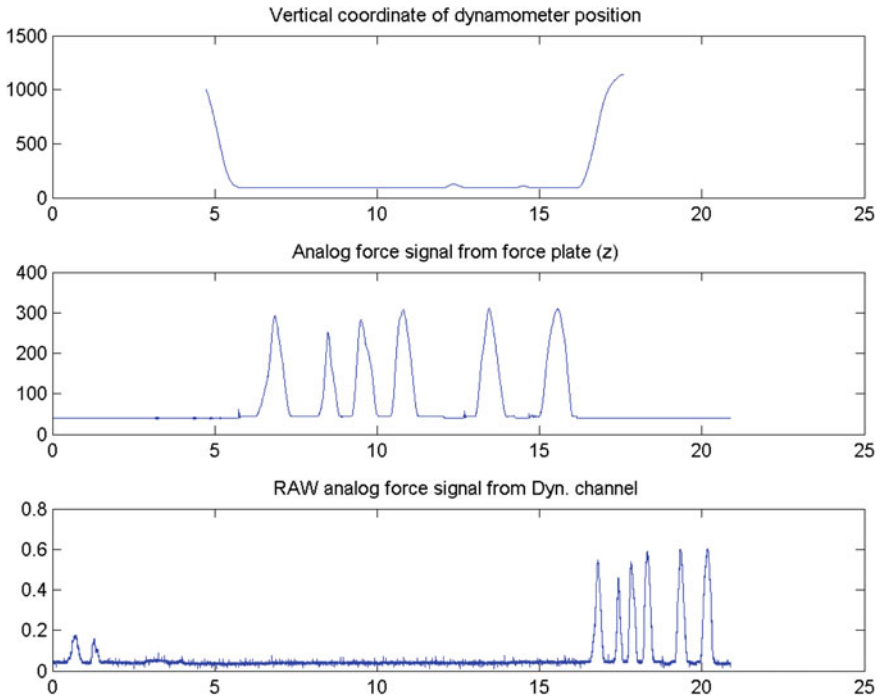


Fig. 2.12 Sync testing of the HHD and Vicon system. Horizontal axis has time, and vertical axes are mm, N, and V, respectively

After an acceptable synchronisation was obtained, the force acquisition campaign was begun: the strength of the knee flexors and knee extensors was measured by means of a clinical protocol defined in close cooperation with the clinical partners of the neuromuscular disease group within the FP7 MD-Paedigree project (Ancillao et al. 2015). The adopted protocol consisted in a ‘make test’ strength measurement method that is widely accepted in the literature (Martin et al. 2006; Wang et al. 2002). The HHD was held by a trained clinician who conducted the trial according to the following directions.

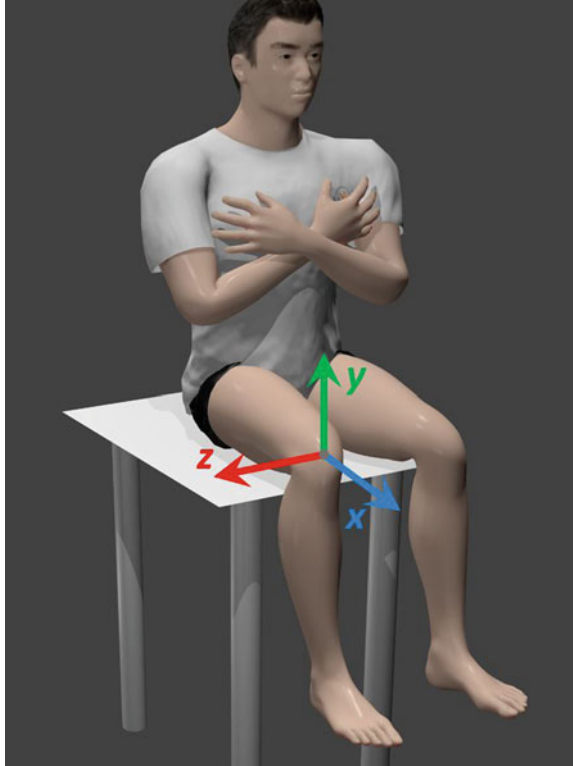
- Extension: Subject sitting, lower legs hanging from the table with hips and knees in flexion at 90° . The HHD was placed proximal to the ankle on the anterior surface of the lower leg. Manual fixation at the thigh and resistance exerted by the operator at the shank in knee flexion direction.
- Flexion: Subject sitting, lower legs hanging from the table with hips and knees in flexion at 90° . The HHD was placed proximal to the ankle on the posterior surface of the lower leg. Manual fixation at the thigh and resistance exerted by the operator at the shank in knee extension direction.

The operator was instructed to counteract the force of the patient by trying to keep the shank still. The patient had to exert maximum force against the HHD for

about five seconds. The participants were also instructed to avoid explosive contraction but to increase force gradually to the maximum (Wang et al. 2002). The distance between the knee rotation centre and the HHD application landmark was manually measured by the operator by using a measuring tape. This distance was needed to compute joint moment. The trials were repeated 5 times for knee extension and 5 times for knee flexion with a resting time of about 30 s between trials to avoid fatigue effects.

Data recorded for each trial were pre-processed by Vicon Nexus 1.7 software (Vicon Motion Systems, UK). Pre-processing included: track labelling, interpolation, smoothing, and C3D export. Advanced data processing was achieved by means of some ad hoc built MATLAB[®] (MathWorks, USA) scripts. Data processing was required to define a local coordinate system (CS) for the shank. The CS of the shank was needed to identify force components and was defined as follows (Fig. 2.13).

Fig. 2.13 Local reference system defined for the knee joint



- *y-axis*: Parallel to the main axis of the shank, directed from the ankle joint to the knee joint
- *yz plane*: The plane containing the y-axis and the vector from the knee centre to the lateral tibial marker. z-axis is in the mediolateral direction, pointing to the right of the subject
- *Origin*: Knee centre, approximated as midpoint between markers on the knee

In accordance with the literature, for each trial we recorded the maximum force measured by the HHD, here addressed as nominal force (F_{nom}), and the nominal knee moment (M_{nom}) obtained simply by multiplying force to the lever arm (Figs. 2.14a, b). Taking advantage of the motion capture data, we also measured the following kinematic and kinetic quantities.

- Knee range of motion (RoM), as the angular displacement of the knee from the resting position (Fig. 2.14c)
- Angles between the dynamometer and the shank on the sagittal plane (A1) and on the transverse plane (A2) at the instant in which the maximum force is recorded (Fig. 2.14d)
- Ranges of motion of the angles A1 and A2 (RoM-A1 and RoM-A2) during the whole trial
- The components of force and moment (F_x , F_y , F_z , M_x , M_y , M_z), accurately computed by knowing the actual position of the HHD and the actual direction of its sensible axis

The main component of knee moment lies on the z-axis. The lateral component ideally should be ~ 0 Nm. Quality analysis of strength measurements is conducted on the moment results, inasmuch as they take into account overall effects due to the positioning and orientation of the HHD performed by the operator.

Some root mean square error (RMSE) coefficients were computed in order to quantify the disagreement between the moment components actually measured (M_x , M_y , M_z) and the nominal value of the moment (M_{nom}). Following its definition, M_{nom} was assumed having effect only on the z-axis, therefore RMSE coefficients were defined, and expressed as percentage of the maximum nominal moment, according to the following equations.

$$\text{RMSE}_z = \sqrt{\frac{\sum_{i=1}^N (M_z^i - M_{\text{nom}}^i)^2}{N}} * \left| \frac{100}{\max(M_{\text{nom}})} \right| \quad (2.1)$$

$$\text{RMSE}_x = \sqrt{\frac{\sum_{i=1}^N (M_x^i)^2}{N}} * \left| \frac{100}{\max(M_{\text{nom}})} \right| \quad (2.2)$$

$$\text{RMSE}_y = \sqrt{\frac{\sum_{i=1}^N (M_y^i)^2}{N}} * \left| \frac{100}{\max(M_{\text{nom}})} \right| \quad (2.3)$$

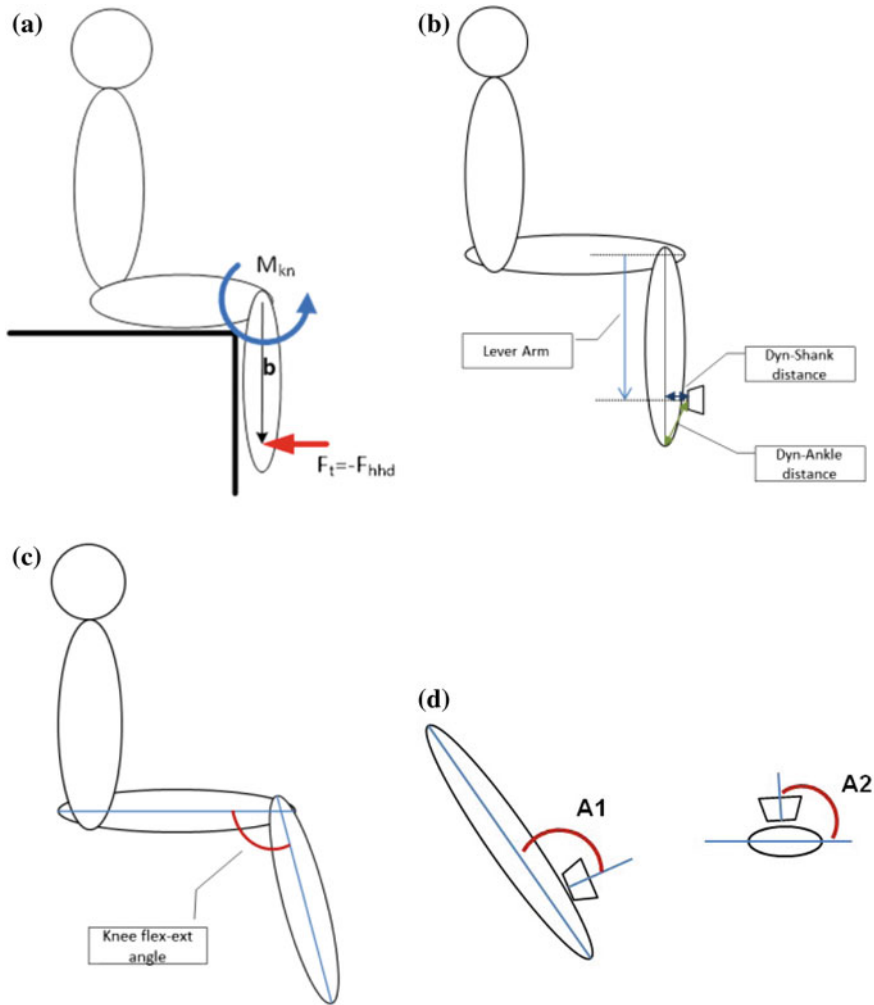


Fig. 2.14 Graphical representation of the measured parameters. **a** Force and moment. **b** Distances. **c** Knee flexion extension angle. **d** HHD positioning angles with respect to the shank

$$RMSE_{xy} = \sqrt{(RMSE_x)^2 + (RMSE_y)^2} \quad (2.4)$$

where N is the number of trials recorded for each subject and $\max(M_{nom})$ is the maximum value of the subject's nominal moment. RMSEs were computed for each subject.

RMSEs of lateral components, that is, x and y , were merged within a single parameter (Eq. (2.4)) that provides an evaluation on the magnitude of lateral components of moment that are neglected in usual clinical strength assessment.

All the parameters were averaged between the five repetitions of each subject. The coefficient of variation (CV) of each parameter, defined as the percentage ratio between standard deviation (SD) and mean was computed to quantify repeatability within the single subject.

$$CV\% = \left| \frac{SD}{\text{mean}} \right| * 100 \quad (2.5)$$

Figure 2.15 shows the force profiles recorded in a preliminary strength trial. In the first figure the distance between the HHD and the shank is visualised. When this distance approaches zero it means that the HHD is in contact with the shank. The green vertical lines represent the trial starting and trial ending events that were used to cut the data tracks. The graphs on the second column show the force profile as measured by the HHD, the force projections on the directions of the CS of the shank, and the moment of the knee along the three axes of the CS. The first column depicts the A1 and A2 angles and knee flexion angle.

Table 2.1 shows the results of kinematic and kinetic analysis of knee extension and flexion trials. The mean value and SD between subjects is reported for each parameter.

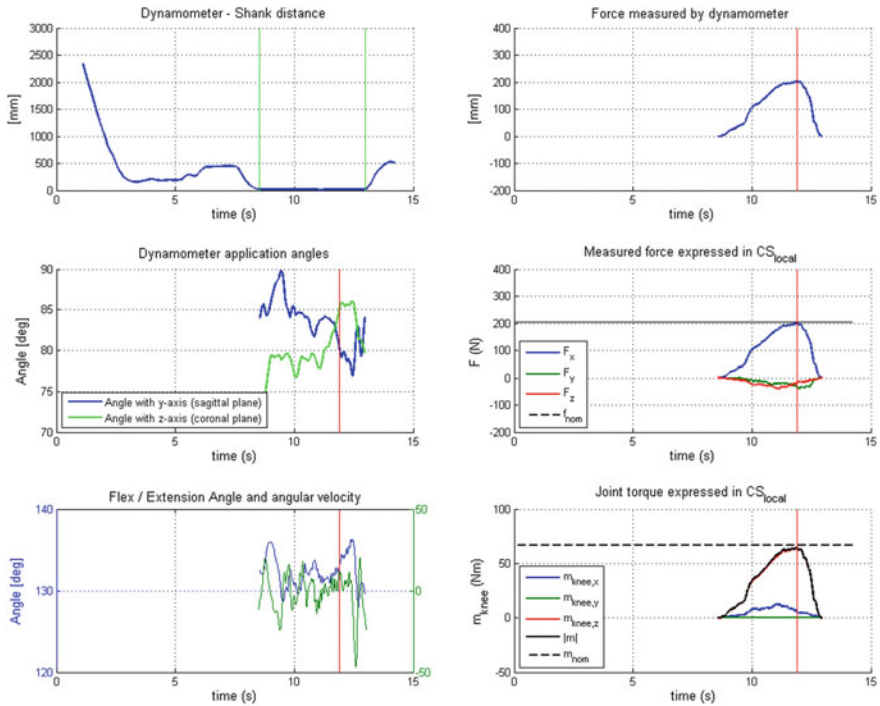


Fig. 2.15 Analysis of forces and position for one strength trial

Table 2.1 Numerical results of the preliminary trials. A1 is the angle on the sagittal plane. A2 is the angle on the transverse plane. F_{nom} is the nominal force. M_{nom} is the nominal moment

Parameter	Knee extension		Knee flexion	
	Mean (SD)	CV% (SD)	Mean (SD)	CV% (SD)
Knee RoM (°)	32 (12)	20.9 (12.1)	27 (5)	16.7 (7.2)
A1 (°)	93 (7)	3.4 (2.0)	87 (8)	9.1 (11.4)
RoM-A1 (°)	14 (6)	27.3 (8.6)	20 (14)	36.3 (14.9)
A2 (°)	90 (8)	5.7 (2.1)	101 (8)	4.1 (1.9)
RoM-A2 (°)	21 (7)	27.1 (9.5)	16 (6)	34.1 (17.1)
F_{nom} (N)	240 (29)	7.6 (2.9)	-138 (22)	6.9 (1.5)
M_{nom} (Nm)	83 (13)	7.6 (3.0)	-46 (6)	6.8 (1.6)
RMSEz (%)	4.8 (1.7)		11.6 (5.7)	
RMSExy (%)	15.3 (7.6)		20.4 (11.5)	
RMSEz (Nm)	4 (2)		5 (3)	
RMSExy (Nm)	12 (5)		9 (6)	

Because the ‘make’ method was adopted for strength measurements (Martin et al. 2006; Wang et al. 2002), the subject’s limb should remain motionless across the trial; that implies a RoM ideally near 0°. In our analysis, the measured angular RoM was never equal to 0° and we observed a value of $32 \pm 12^\circ$ across the subjects; hence, the operator was not able to hold the limb completely still. This fact affected the strength measurement, as the operator was not able to exert a correct opposing force to the subject. This result is consistent with the results of Kim et al. (2014) that proved a better measurement validity when the HHD is fixed with Velcro than when it is held by the operator.

From the preliminary results, we observed that Knee RoM had more variability in the knee extension trials than knee flexion, in terms of both mean value and CV. This finding can be related to the lower force exerted in flexion trials and it was in agreement with the results of Laing et al. (1995).

Repeatability of measurements was quantified for each subject by means of the CV % of the force and moment. The average CV% was low (<10%) for both knee extension and knee flexion trials, meaning high repeatability of measurements, consistent with previous studies (Kim et al. 2014; Martin et al. 2006; Phillips et al. 2000).

Positioning angles A1 and A2 were close to 90° with a low CV% (<10%), meaning a good positioning of the HHD on the shank. The worst results were obtained for the knee flexion trials, due to the operator’s position that had to extend his arm behind the shank and therefore gave him poor control of positioning.

Ranges of motion of angles A1 and A2 quantified the stability of the operator’s hand across the trial. They were in the range of 20° with high CV%, >25% for knee extension, and >30% for knee flexion, meaning instability and movement of the HHD during the trial. This finding confirmed the operator’s dependence on the measurement quality (Bohannon 1986; Kim et al. 2014; Martin et al. 2006; Willemse et al. 2013) and also confirmed the poor control of HHD positioning in knee flex trials.

In a clinical context, only nominal force and moment are measured and they are assumed as the force and moment on the axis of flex/extension and lateral components are neglected. RMSE parameters are therefore computed in order to quantify the inaccuracy committed by neglecting the lateral components. More precisely, $RMSE_z$ represented the error between the nominal moment and the component of the actual moment on the z -axis, that is, the flex/ext axis, and $RMSE_{xy}$ represented the magnitude of lateral components of the actual moment.

The knee extension trials had a $RMSE_z < 5\%$ indicating a low error, whereas $RMSE_z$ for flexion was $>10\%$. Also $RMSE_{xy}$, that represents the effect of lateral components of the moment, was higher for knee flexion (20.4%) than knee extension (15.3%).

Absolute RMSEs on the main axis were comparable with the respective uncertainty level. RMSEs of lateral components were slightly higher but lower than the moment on the main axis. These findings were connected to the angular displacement observed in A1 and A2 values and confirmed that angular misplacement induced an inaccuracy in the estimation of flex/extension moment.

The force and the moment exerted by the subjects were higher in knee extension trials with respect to knee flexion trials. Knee flexion trials had some issues due to HHD positioning. These issues were represented by angles A1 and A2 and their ranges of motion. Moreover, the operator was not able to keep the subject's limb perfectly still. Therefore, specific attention has to be paid to HHD positioning in knee flexion and extension trials. Stability of the HHD is crucial and therefore training of the operator is extremely important. Moreover, the operator should be strong enough to exert a force equal to the force produced by the subject and avoid motion of the limb (Ancillao et al. 2015).

The preliminary analysis proved that the motion capture protocol was robust and able to reconstruct the kinematics.

The main limitations observed in the preliminary trials were:

- The HHD used was a 1-component load cell. A multicomponent load cell would be useful to evaluate the effect of lateral components of force and moment better.
- Need to simplify the system by removing the LabVIEW software running on a dedicated machine and the D/A board, having the whole system running on the same machine.
- Manual synchronisation of data was time consuming. Synchronising data at the time of recording would dramatically reduce processing time.
- Other anatomical districts should be taken into account by the protocol.
- Limited number of subjects.
- Need of software to batch-process a large amount of data with reduced user action.
- Need to 'summarise' results in a few parameters and make them simple to understand for clinicians.

2.3 Final Experimental Setup

To overcome the limitations observed in the preliminary analysis, the materials and methods of the research were modified as follows (Ancillao et al. 2015, 2017a, b).

2.3.1 Subjects

Thirty healthy adult subjects were enrolled in the study: 18 males, 12 females, age 26.2 ± 2.1 years, height 173.6 ± 7.2 cm, body mass 68.1 ± 8.7 kg. Subjects must not have had any neurological or orthopaedic disorder and must not have undergone surgery to the lower limb joints. All the subjects were evaluated by a physiatrist before the trials. All the subjects were right-handed even though this was not an inclusion criterion. This study complied with the principles of the Declaration of Helsinki.

2.3.2 Instrumentation

In order to overcome the limitations observed for the one-component HHD (as used for the preliminary analysis) a multicomponent analog load cell was used. Specifically, the limitations of a one-component HHD were identified as (i) data synchronisation issues, (ii) loss of information about the lateral components of force, and (iii) no direct measurement of torque.

The sensor chosen was a six-component HHD based on a Gamma(r) F/T load cell (ATI Industrial Automation, USA), shown in Fig. 2.16. Measurement ranges and metrological characteristics are shown in Tables 2.2 and 2.3. This model was chosen because the sensing range on the z -axis was compatible with the range of strength measurements to be conducted. Moreover, the physical dimensions (Table 2.3) were comparable to the MicroFET2(r) dynamometer, therefore the load cell could be held by hand and used as an HHD.

The load cell was equipped with an ad hoc designed aluminium force-transferring layer and a foam layer on the side that was in contact with the subject's limb in order to maximise the subject's comfort during the measurement. Mass of the load cell was 0.255 kg, diameter 75.4 mm, and height 33.3 mm. Geometrical details of the layers are shown in Figs. 2.17 and 2.18.

The load cell was connected through a shielded cable to a pre-amplifier box, powered by 5 V DC current. Downstream of the pre-amplifier box, the analog voltage output of each channel was available. Each channel delivered its output by two coupled wires protected by a shielded cable. A total of 12 wires plus ground and shield were connected to the analog input box of the Vicon system and acquired in differential mode (Fig. 2.19). By using this method of data acquisition, analog data from the load cell were synced to the kinematic data by the internal triggering of the Vicon system.

Fig. 2.16 ATI Gamma
6-component F/T Sensor



Table 2.2 Ranges and resolution of the ATI Gamma F/T Sensor

Sensing ranges			
Fx, Fy	Fz	Tx, Ty	Tz
130 N	400 N	10 Nm	10 Nm
Resolution			
Fx, Fy	Fz	Tx, Ty	Tz
1/40 N	1/20 N	1/800 Nm	1/800 Nm

Table 2.3 Technical specifications and metrological characteristics of the ATI Gamma F/T Sensor

Single-axis overload	
Fx, Fy	±1200 N
Fz	±4100 N
Tx, Ty	±79 Nm
Tz	±82 Nm
Stiffness	
X-axis and Y-axis forces (Kx, Ky)	9.1×10^6 N/m
Z-axis force (Kz)	1.8×10^7 N/m
X-axis and Y-axis torque (Ktx, Kty)	1.1×10^4 Nm/rad
Z-axis torque (Ktz)	1.6×10^4 Nm/rad
Resonant Frequency	
Fx, Fy, Tz	1400 Hz
Fz, Tx, Ty	2000 Hz
Physical Specifications	
Weight*	0.255 kg
Diameter*	75.4 mm
Height*	33.3 mm

*Specifications include standard interface plates

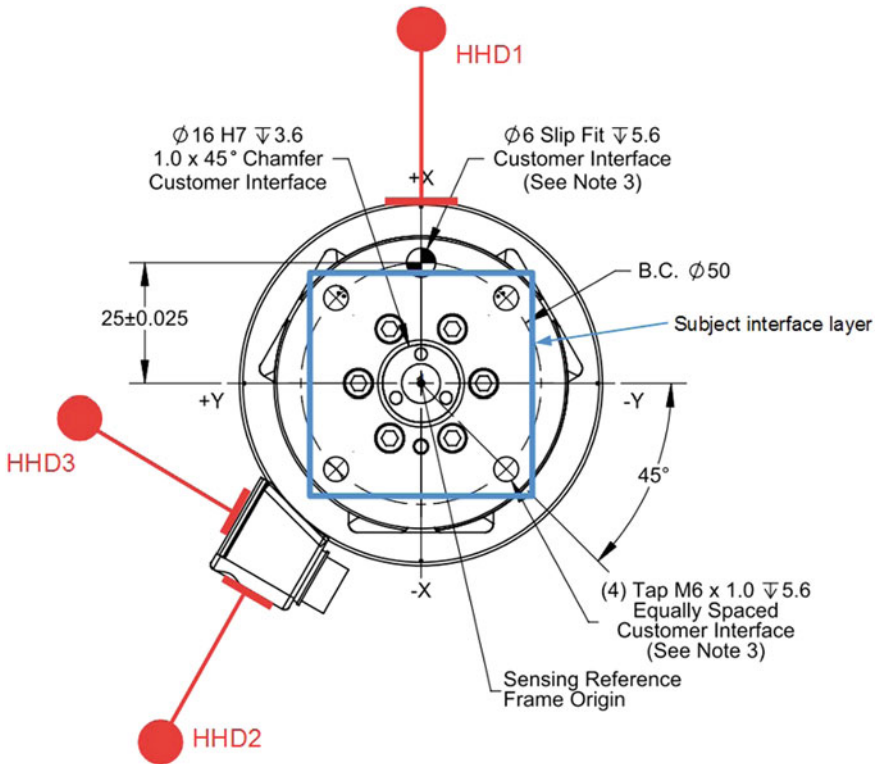


Fig. 2.17 Geometrical specifications of the ATI Gamma F/T Sensor and application of the Vicor markers (red)

The load cell and the acquisition system were tested by applying a known load to the cell and the force platform in series (Fig. 2.20). Results of this test showed the proper operation of the measurement chain.

The motion was recorded by means of a Vicon MX Optoelectronic System (Oxford Metrics, UK), equipped with eight cameras. The OS was able to reconstruct the x , y , z coordinates of the markers and their motion in a 3D virtual environment.

Static and dynamic calibration tests, performed in accordance with the manufacturer's indications, were conducted before each participant's trial session and they showed that overall RMS error of marker coordinates in three-dimensional space was less than 1 mm in a calibrated volume of about $2 \times 1 \times 2 \text{ m}^3$.

Calibration allowed building a local reference system for each camera and defining the position in space of each camera with respect to an absolute reference system. The calibration procedure adopted by the Vicon MX system is called *DynaCal3*. It uses a T-shaped wand, with markers on it, for both dynamic and static calibration procedures. Markers are placed in such a way as to determine univocally the wand's position and orientation in space.

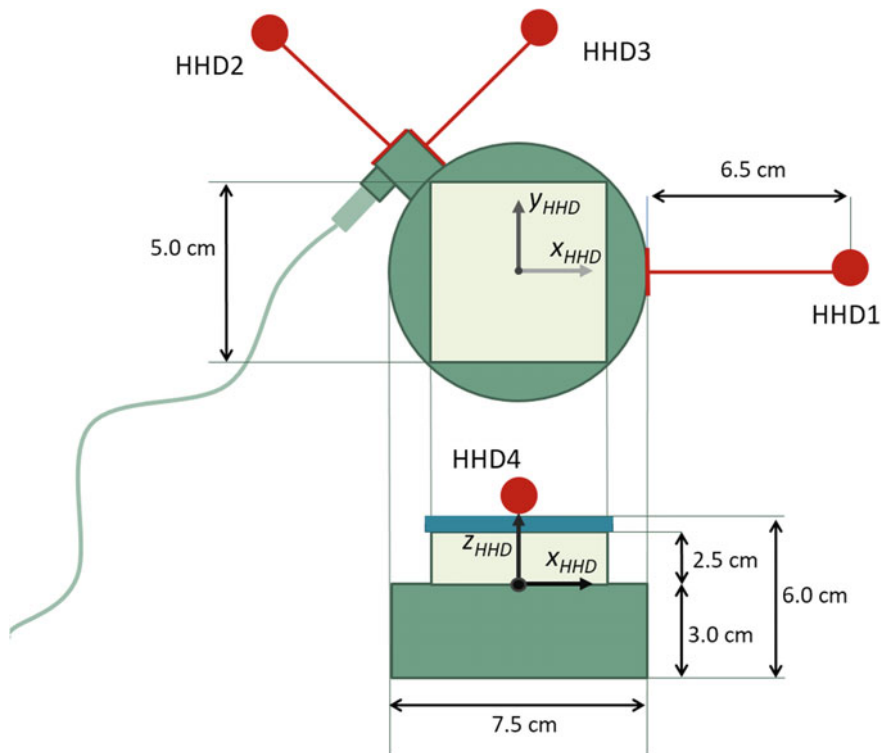


Fig. 2.18 Design of force transferring layers and their dimensions. Landmarks for Vicor markers are visualised in red. The local reference system is also indicated

Fig. 2.19 Vicor analog junction box and connection of the 6 channels and ground (13 connectors) from the load cell

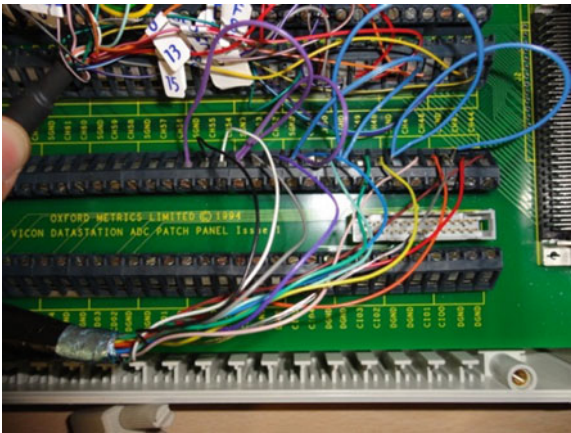


Fig. 2.20 Testing of the load cell by applying a known load in series to the load cell and the force plate



Dynamic calibration consists in randomly moving the wand along the volume to be calibrated, paying attention to have each camera recording an adequate amount of frames containing the full visible wand. This procedure allows defining the boundaries of the acquisition volume and computation of camera parameters and relative position. This is made possible by comparing the measured distance between the wand's markers and their actual known distance. Static calibration consists in placing the wand on the floor and recording its position. This allows defining the absolute reference system of the lab and to compute the position of each camera with respect to this reference system. The crossing point of the wand's arm becomes the origin of the Cartesian space, and the arms define the axes. By convention, the z -axis is defined pointing upwards from the floor.

Vicon Nexus software allowed acquiring and processing the system calibration, to view in real-time the acquisition volume as well as the output of each camera. At the end of the recording, the software could reconstruct the x , y , z coordinates of the markers, allowing their labelling and construction of the biomechanical model.

The signal produced by the load cell was recorded and synchronised to the kinematic data and to the force platforms of the lab. The signal was stored within the same file container of kinematic and force platform data, that is, C3D format.

2.3.3 Motion Capture Protocol

The six-component HHD was equipped with four passive markers as shown in Fig. 2.21. Markers were placed on sticks rigidly fixed to the HHD to avoid covering by the operator's hand. The central marker was placed in the midpoint of the patient-interface area of the HHD needed to locate its centre with respect to the other markers that were used to build a local reference system (Figs. 2.18 and 2.21). The central marker was removed during the measurement of strength and its

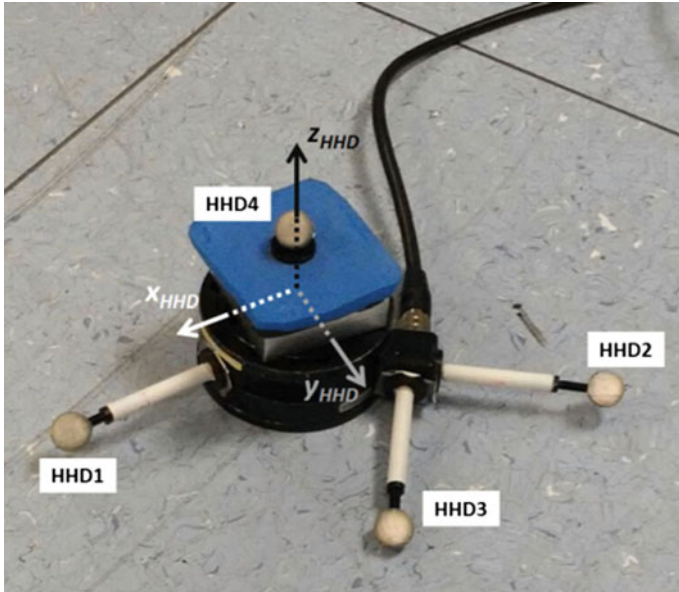


Fig. 2.21 Six-component load cell equipped with force transferring layers. Markers are identified by their names and the local reference system is represented

position was reconstructed by using a nonoptimal localisation procedure based on the three fixed markers (Ancillao et al. 2013).

Markers on the subjects were placed according to an ad hoc defined marker protocol composed of 26 markers placed on the subject's skin surface (Fig. 2.22). Landmarks were identified as posterior and anterior iliac spines (4 markers), lateral thighs (2 markers), lateral and medial epicondyles (4 markers), lateral and medial malleoli (4 markers), lateral shanks (2 markers), second metatarsal head (2 markers), and calcaneus (2 markers). Finally, two clusters of three markers (6 markers) were applied on the thighs (in correspondence with the quadriceps femoris).

Positions of knee and ankle centres were computed as the midpoint between the two markers on epicondyles and on the malleoli, respectively. The hip centre was reconstructed solving the Plug-in-Gait model, which is a modified version of the Davis protocol (Davis et al. 1991), when the subject is in the upright position. Instead, the position of the hip centre when the subject assumed the seated position, was reconstructed by means of an optimal localisation procedure (Cappozzo et al. 1997) using the markers on the thigh, inasmuch as the markers on the pelvis were not always visible to the OS cameras.

To record ankle plantarflexion and dorsiflexion strength trials, a modified version of the marker protocol was used instead (Fig. 2.23). This adaptation was needed to allow the subject to assume the testing position, that is, lying on the bed (Ancillao et al. 2017a). In ankle strength trials, landmarks were identified as lateral and medial femoral epicondyles (4 markers), lateral and medial malleoli

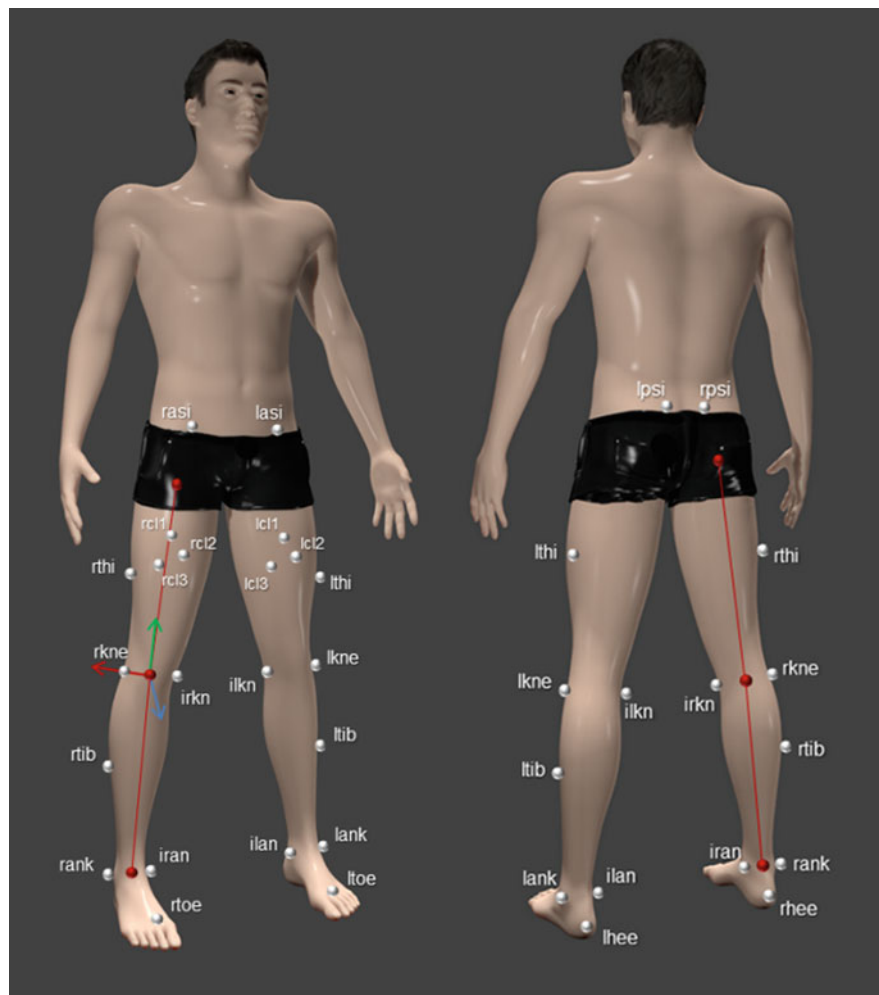


Fig. 2.22 The marker protocol used to reconstruct the position of subject’s lower limb. White dots represent the markers used with the optoelectronic system; red dots are the computed joint centres (virtual markers). Local reference system of the shank is shown (Ancillao et al. 2017b)

(4 markers), lateral shanks (2 markers), head of first metatarsal (2 markers), and head of fifth metatarsal (2 markers). The left leg was included as well in the protocol design, in order to enable recording and processing of strength trials in left-handed subject.

Before recording the strength trials, the recording of a calibration (static) trial for each subject was needed. In the static trial, the subject was asked to stand up motionless in the calibrated volume for about five seconds while wearing the full marker set, as in Fig. 2.22. At the same time the HHD was kept still on the laboratory floor within the calibrated volume of the OS, while no load was applied.



Fig. 2.23 Graphical representation of the marker protocol used in ankle strength trials

The static trial was necessary to evaluate: (i) the position and the orientation of the local reference systems identifying the body segments and the HHD, and (ii) the offset signals gathered by the HHD.

2.3.4 *Strength Protocol*

Strength of the knee flexors and knee extensor muscles was measured using a clinical protocol defined in close cooperation with the clinical partners of the neuromuscular disease group within the FP7 MD-Paedigree project. This protocol consisted in a ‘make test’ method (Bohannon 1988; Laing et al. 1995; Martin et al. 2006; Wuang et al. 2013).

The HHD was operated by a trained clinician who performed the trials in accordance with (Eek et al. 2006): subjects were sitting on a bench, lower legs were hanging with hips and knees in flexion at about 90°. They were stabilised to reduce compensation from other muscle groups and the movements of the thigh were impeded by a belt connected to the bench. The HHD was placed proximally to the ankle, on the anterior surface of the lower leg for the knee extension movement (Fig. 2.24), and on the posterior surface of the lower leg for the knee flexion movement (Fig. 2.25). The operator was expected to assume a comfortable position that allowed him to exert a properly calibrated force in the opposite direction to that of the patient’s (Fig. 2.26).

Fig. 2.24 HHD placement
for knee extension trials



Fig. 2.25 HHD placement
for knee flexion trials

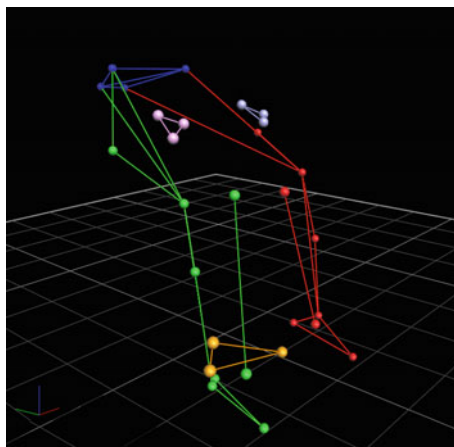


The markers that were placed on the subject and on the HHD allowed obtaining a biomechanical model able numerically to reconstruct the position and motion of the subject, the operator, and the HHD (Fig. 2.27). Such information was used to compute the parameters of interest.

Fig. 2.26 Graphical rendering of the correct positioning for a knee extension trial



Fig. 2.27 Biomechanical model of the lower limb with the HHD (yellow triangle) placed in proximity to the right ankle



The subjects were instructed to exert their maximal force against the HHD for about five seconds while the operator counteracted the force trying to keep the shank still. The participants were also instructed to avoid explosive contraction but they were invited to increase force gradually from zero to the maximum achievable value (Wuang et al. 2013). Participants were tested individually by a single operator. Trials were repeated five times for both knee extension and knee flexion with a resting time of about 30 s between trials to avoid fatigue effects in both subject and operator. The session for each participant lasted approximately 30 min.

As for knee flexion/extension, the strength of the ankle dorsiflexor and plantarflexor muscles was estimated by using a previously validated clinical protocol (Eek et al. 2006) consisting in a ‘make test’ method (Bohannon 1988; Laing et al. 1995) in agreement with the clinical partners of the neuromuscular disease group

within the FP7 MD-Paedigree project. In both ankle plantarflexion and dorsiflexion trials, the subject was lying on the bed with the ankle in supine position and the marker set applied. HHD was placed under the foot sole in the metatarsal region for plantarflexion testing (Fig. 2.28), and on the upper metatarsal for dorsiflexion testing (Fig. 2.29). The subjects were instructed to push against the HHD exerting their maximum force while the operator had to counteract the patient's force to keep the foot still for about five seconds. The participants were instructed to avoid explosive contraction but to increase force gradually from zero to the maximum achievable value (Wuang et al. 2013). Trials were repeated five times for both plantarflexion and dorsiflexion with a resting time of about 30 s between trials to avoid fatigue effects in both subject and operator.

Fig. 2.28 HHD placement for ankle plantarflexion trials

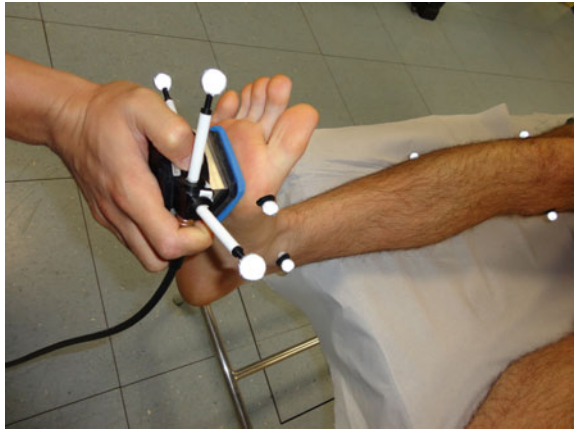


Fig. 2.29 HHD placement for ankle dorsiflexion trials



2.4 Data Processing

The kinematic data and the force signal from the load cell were recorded simultaneously and stored within the same data file, that is, a C3D file that served as a container for motion capture data.

Pre-processing was done in Vicon Nexus 1.7 (Oxford Metrics, UK). It included: signal denoising, track labelling, artefact removal, and data compression. The pre-processed data for each trial were stored within the same C3D files. Further data processing was achieved by means of ad hoc designed algorithms implemented in MATLAB (MathWorks, USA). C3D data files were then imported within the MATLAB environment by means of ad hoc designed libraries, and then processed as explained in the following.

The overall flowchart of data processing is depicted in Fig. 2.30. Calibration (a.k.a. static) trials were processed first. The static parameters were required by the dynamic processing script, namely the dynamic engine. The purpose of the dynamic engine was to interpret motion capture and HHD data in order to obtain some kinematic and kinetic indices that were used to describe the quality of strength measurements. Static processing and dynamic processing are explained through Figs. 2.31 and 2.32.

The first step, required to compute the biomechanical parameters, was the definition of some anatomical local reference systems (LRS) for body segments whose position in space needed to be identified. LRS were designed on the basis of external markers placed on the skin of the subject. For the knee flexion/extension trials, two LRS were defined: for the shank, namely LRS_{SH} (Fig. 2.33), and for the HHD, namely LRS_{HHD} (Fig. 2.21). For the ankle strength trials, in addition to the previously defined LRS, a LRS for the foot, namely LRS_{FT} , was also defined.

LRS_{SH} was defined as

- y_{SH} , the unit vector from ankle centre to knee centre, directed upwards
- x_{SH} , the unit vector perpendicular to the plane defined by the knee medial and lateral epicondyles and the ankle centre, and pointing forwards
- z_{SH} , the unit vector perpendicular to x_{SH} and y_{SH}
- Origin, located at the knee centre

LRS_{HHD} was defined as

- $vmkr$, the virtual marker defined as the projection of HHD4 on the plane represented by HHD1, HHD2, and HHD3
- x_{HHD} , the unit vector from $vmkr$ to HHD1
- z_{HHD} , the unit vector perpendicular to the plane defined by HHD1, HHD2, and HHD3, pointing upwards
- y_{HHD} , defined as the cross-product between z_{HHD} and x_{HHD}
- Origin, the virtual marker on the line between $vmkr$ and HHD4 with an offset from HHD4 of 2.7 cm, that is, the sum of thickness of the force coupling layers and the marker's radius

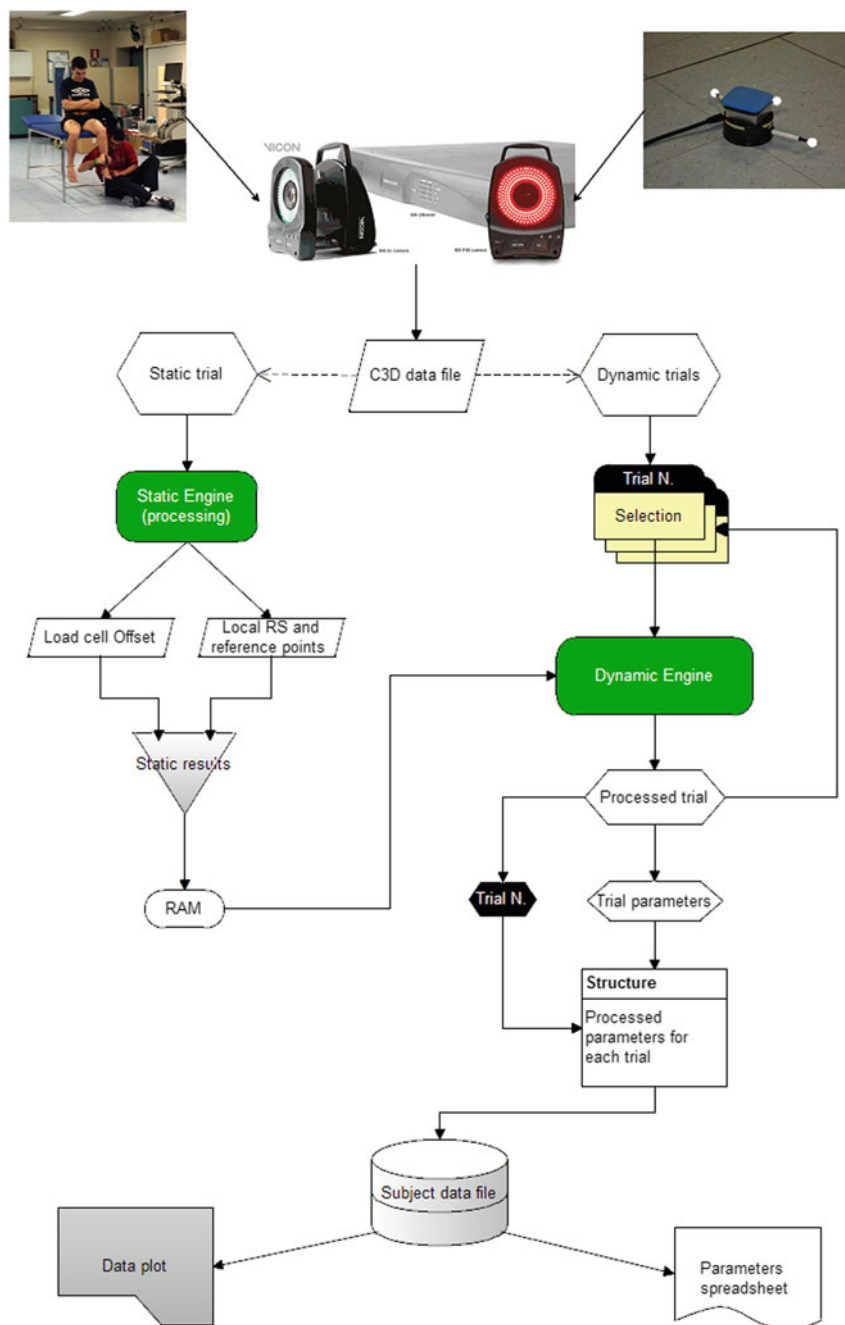


Fig. 2.30 Overall data-processing flowchart

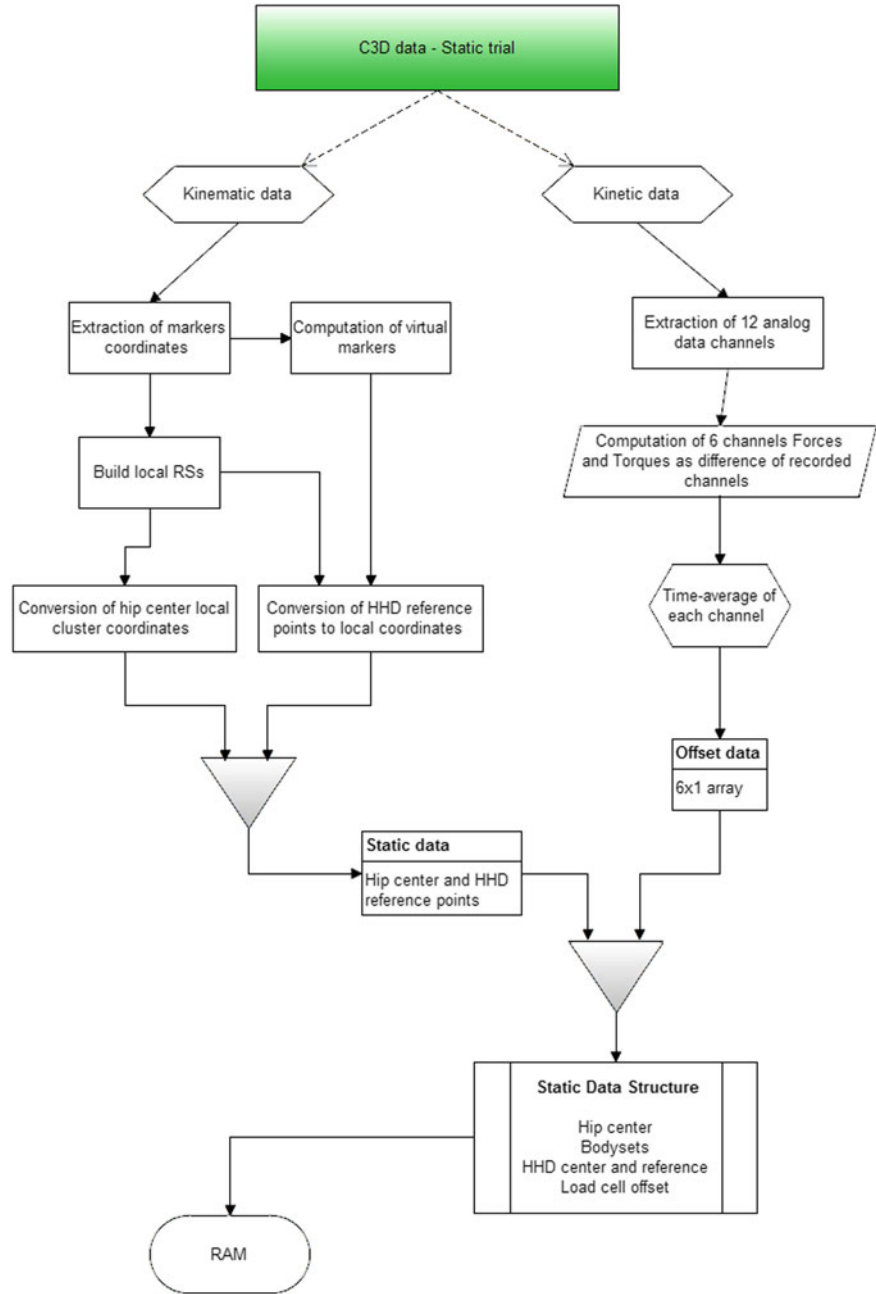


Fig. 2.31 Flowchart for the static processing script

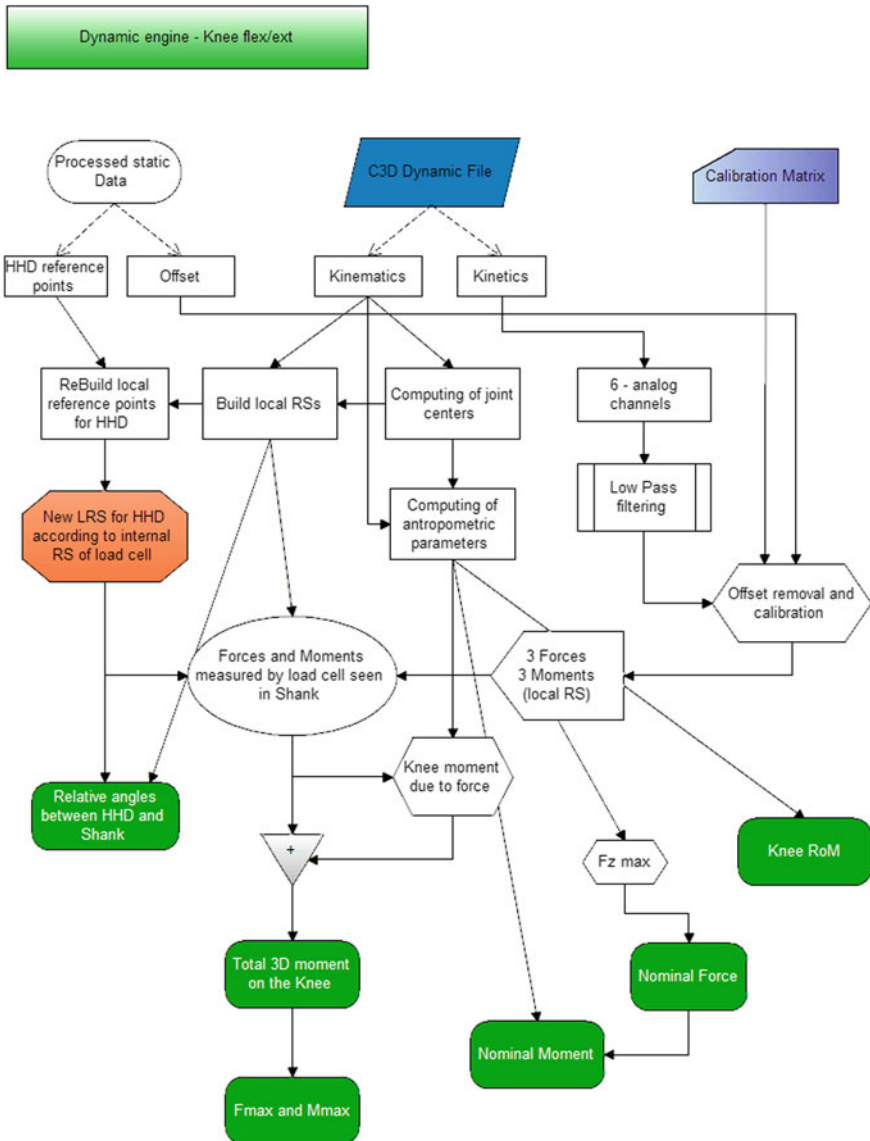


Fig. 2.32 Flowchart for the dynamic-processing script

LRS_{FT} was defined as

- y_{FT} : unit vector from ankle centre to knee centre
- z_{FT} : unit vector perpendicular to the plane defined by the knee centre, the ankle centre, and midpoint between markers on the first and fifth metatarsal, pointing to lateral direction

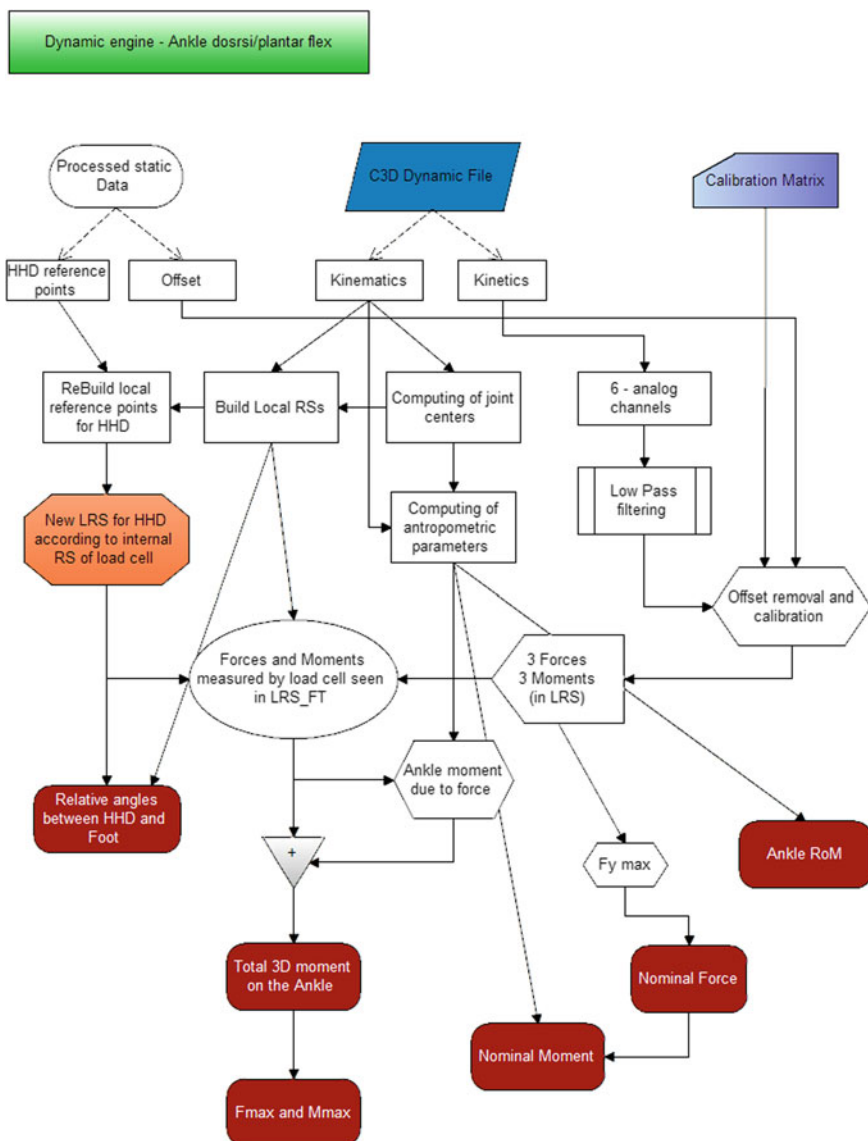


Fig. 2.33 Flowchart for the dynamic-processing script

- x_{FT} : defined as cross-product between y_{FT} and z_{FT} axes
- Origin: ankle centre

The LRS allowed localising in space the anatomical segments modelled as rigid bodies and computing Euler's angles between those bodies. A set of parameters describing the motion of the limb and HHD placement was obtained. Such computed parameters were classified as 'kinematic' indices and 'kinetic' indices.

2.4.1 Indices for Knee Strength Trials

The kinematic indices describing the knee trials were computed for both knee extension and knee flexion and they were averaged between the five repetitions of each subject (Ancillao et al. 2017b). Such indices were defined as

- Range of motion (RoM) of knee angle, defined as the difference between the maximum and minimum of knee angle measured throughout the trial. Specifically, the knee angle was computed as the angle between two vectors (Vimercati et al. 2013), namely the vector from knee centre to hip centre and the vector from knee centre to ankle centre. RoM can be assumed as representative of the quality of strength measurements as the limb should ideally remain still during the trial. Therefore low values of RoM represent higher adherence to the selected protocol.
- A1 and A2, that represent the angles between z_{HHD} and y_{SH} and between z_{HHD} and z_{SH} , respectively (Figs. 2.34 and 2.35). A1 and A2 were evaluated at the instant in which the measurement of strength was gathered, that is, when the maximal force was recorded. A1 and A2 should be ideally equal to 90° and their deviations from this value, namely δA_1 and δA_2 , represent the incorrect positioning of HHD on the sagittal and horizontal planes, respectively.

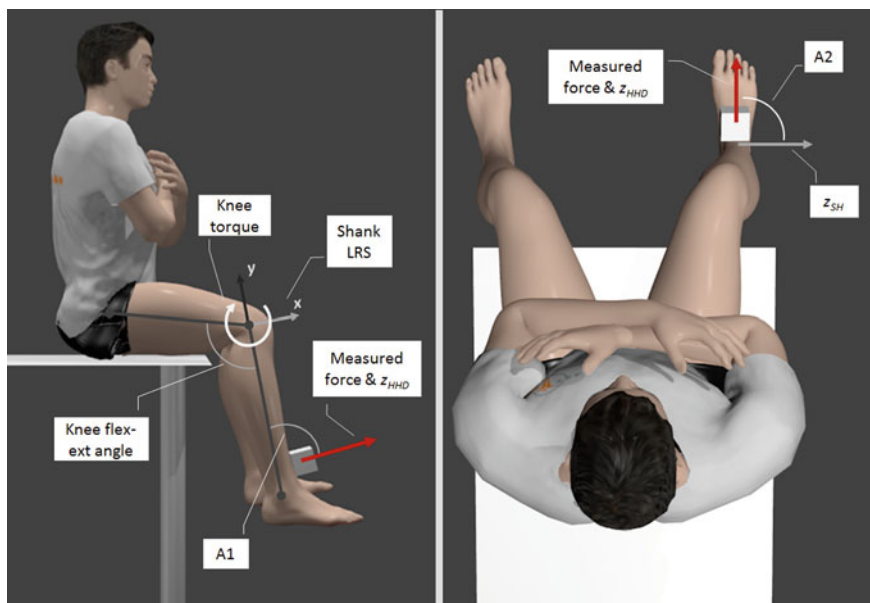


Fig. 2.34 3D rendering representing subject's position, local reference systems, and computed parameters. Lateral view and top view (Ancillao et al. 2017b)

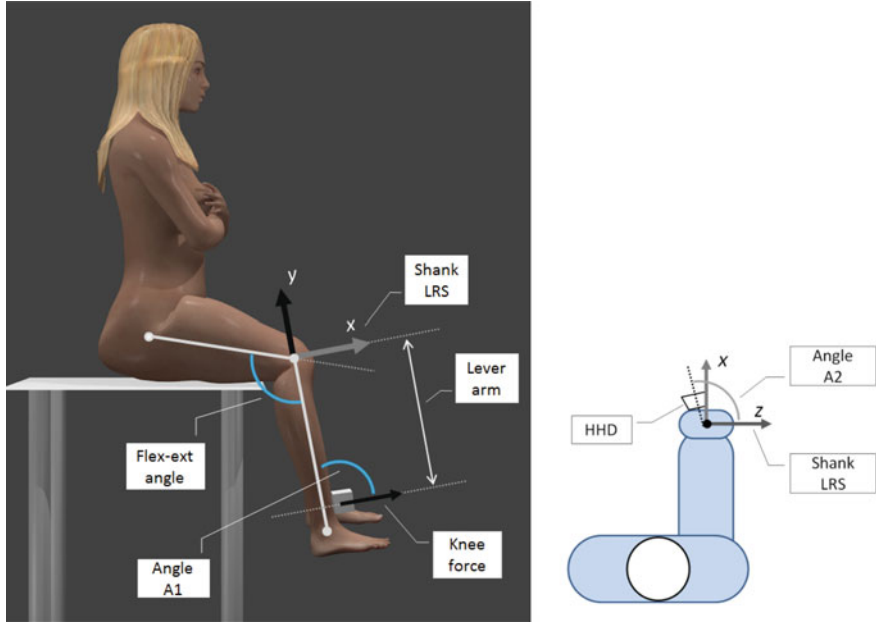


Fig. 2.35 Alternative rendering and detailed drawing explaining the definition of A1 and A2 angles, knee angle, and lever arm

In order to assess repeatability of kinematic measurements, the coefficient of variation was computed for each parameter and it was addressed as CV_{RoM} , $CV_{\delta A_1}$, and $CV_{\delta A_2}$. The CV was defined as the percentage ratio between the standard deviation (SD) and the mean value across the five repetitions for each subject.

Some kinetic parameters were computed in order to describe forces and moments acting on the knee joint centre. Force and moment were represented in LRS_{SH} (${}^{SH}\mathbf{F}$ and ${}^{SH}\mathbf{M}$, respectively):

$${}^{SH}\mathbf{F} = {}^{SH}\mathbf{R}_{HHD} \cdot {}^{HHD}\mathbf{F} \quad (2.6)$$

$${}^{SH}\mathbf{M} = {}^{SH}\mathbf{R}_{HHD} \cdot {}^{HHD}\mathbf{M} + {}^{SH}\mathbf{o}_{HHD} \times {}^{SH}\mathbf{F} \quad (2.7)$$

where ${}^{HHD}\mathbf{F}$ and ${}^{HHD}\mathbf{M}$ are the outputs of the HHD, ${}^{SH}\mathbf{R}_{HHD}$ is the rotational matrix which rotates vectors from coordinate system LRS_{HHD} to LRS_{SH} , and ${}^{SH}\mathbf{o}_{HHD}$ is the origin of LRS_{HHD} represented in LRS_{SH} .

The kinetic indices for the knee flexion/extension trials were (Ancillao et al. 2017b):

- F_M , defined as the maximum value of $^{SH}F_y$, which represents the strength measurement.
- F_T , the transverse component of the force exerted by the subject. It allowed the quantification of the intensity of lateral force that cannot be acquired with the usual clinical measurements conducted with a uniaxial load cell:

$$F_T = \sqrt{^{SH}F_y^2 + ^{SH}F_z^2} \quad (2.8)$$

- M_M , defined as the maximum value of the knee flexion-extension moment, that is, $^{SH}M_z$, when the strength measurement is conducted.
- M_T , the transverse component of the knee moment. It represents the moment components that are lost if a uniaxial load cell were used to measure the knee moment:

$$M_T = \sqrt{^{SH}M_x^2 + ^{SH}M_y^2} \quad (2.9)$$

All the computed indices were referred to the time instant when the maximum force was acquired and they were averaged between the five repetitions for each subject. Moreover, each kinetic index was computed for both knee extension and knee flexion trials.

In order to estimate the operator-dependent inaccuracy, the nominal knee strength ($\hat{F}$) and the nominal knee moment ($\hat{M}$) were estimated as they are usually measured in clinical routine. Specifically, clinicians evaluate the force exerted by subjects ($\hat{F}$) by using a uniaxial HHD and subsequently they estimate knee moment multiplying $\hat{F}$ by the distance between the HHD and the knee centre usually determined with a tape measure. Thus, we simulated a uniaxial HHD by focusing only on the $^{HHD}F_z$ force component. In addition, we considered the quantity $^{SH}\mathbf{o}_{HHD}$, that is, the distance between the knee epicondyle and the HHD, as the nominal lever arm for the evaluation of knee moment (Eqs. (2.10) and (2.11)).

$$\hat{F} = ^{HHD}F_z \quad (2.10)$$

$$\hat{M} = ^{SH}\mathbf{o}_{HHD} \cdot \hat{F} \quad (2.11)$$

The inaccuracies of the strength and the knee moment were evaluated as the root mean square error of the differences between the actual values (F_M and M_M) and the nominal ones ($\hat{F}$ and $\hat{M}$); RMSEs were also normalised to the maximum values of $\hat{F}$ and $\hat{M}$ (Eq. (2.12)).

$$\begin{aligned}
 \text{RMSE}_F &= \sqrt{\frac{\sum_{i=1}^N (F_M^i - \hat{F}_i)^2}{N}} \cdot \frac{100}{\max_i(\hat{F}_i)} (\%) \\
 \text{RMSE}_M &= \sqrt{\frac{\sum_{i=1}^N (M_M^i - \hat{M}_i)^2}{N}} \cdot \frac{100}{\max_i(\hat{M}_i)} (\%)
 \end{aligned} \tag{2.12}$$

where N is the number of repetitions of the trials. Thus, RMSE values permitted the overall quantification of inaccuracy occurring in knee strength measurements performed in the clinical routine. To assess the repeatability, the coefficient of variation for $\hat{F}$ and $\hat{M}$ was also computed and addressed as $CV_{\hat{F}}$ and $CV_{\hat{M}}$.

Finally, to give an overall quantification of the quality of strength measurement, a novel synthetic quality index Q_{index} was defined. It was designed in order to take into account both the angular displacements of the HHD (δA_1 and δA_2) expressed as a percentage of 90° , and the transverse component of moment (M_T) expressed as a percentage of the maximum value of the knee moment (M_M). The transverse component of moment was assumed representative of the quality of the measurement because it takes into account both the effects induced by the HHD incorrect positioning and the transversal force components.

$$Q_{\text{index}} = 100 \left(1 - \sqrt{\left(\frac{\delta A_1}{90} \right)^2 + \left(\frac{\delta A_2}{90} \right)^2 + \left(\frac{M_T}{M_M} \right)^2} \right) \tag{2.13}$$

An ideal knee strength measurement implies δA_1 , δA_2 , and M_T equal to zero, that is, Q_{index} equal to 100%. Thus, Q_{index} values lower than 100% indicate a worsening of the strength measurements.

2.4.2 Indices for Ankle Strength Trials

The kinematic indices describing the ankle strength trials were computed for both ankle plantarflexion and ankle dorsiflexion trials and they were averaged between the five repetitions of each subject. Such indices were defined as

- The range of motion (RoM) of the ankle dorsiplantar flexion angle (Fig. 2.34), defined as the difference between the maximum and minimum of angle measured throughout the trial. Ankle angle was computed on the basis of a three-point procedure between knee centre, ankle centre, and the midpoint between markers on the first and fifth metatarsal. As the ankle should ideally remain motionless during the strength measurement, RoM was assumed as a quality indicator of strength measurements: a lower RoM indicates a higher quality of measurement.

- The angles between the HHD z -axis and the transverse and sagittal planes of the foot, namely A_1 and A_2 . A_1 and A_2 were evaluated when the maximum force from the HHD was recorded. Their deviations (i.e. δA_1 and δA_2) from their ideal values (90°) indicate wrong positioning of the HHD during the strength measurement. In the ideal case $\delta A_1 = \delta A_2 = 0^\circ$.

To assess repeatability of the measurements, the coefficient of variation was computed for each parameter and it was addressed as CV_{RoM} , $CV_{\delta A_1}$, and $CV_{\delta A_2}$. The CV was defined as the percentage ratio between the standard deviation (SD) and the mean value across the five repetitions for each subject (Fig. 2.36).

Kinetic analysis was conducted in terms of forces and moments acting on the ankle joint. Forces and moments were expressed in the LRS of the foot (${}^{FT}\mathbf{F}$ and ${}^{FT}\mathbf{M}$):

$${}^{FT}\mathbf{F} = {}^{FT}\mathbf{R}_{HHD} \cdot {}^{HHD}\mathbf{F} \quad (2.14)$$

$${}^{FT}\mathbf{M} = {}^{FT}\mathbf{R}_{HHD} \cdot {}^{HHD}\mathbf{M} + {}^{FT}\mathbf{o}_{HHD} \times {}^{FT}\mathbf{F} \quad (2.15)$$

where ${}^{HHD}\mathbf{F}$ and ${}^{HHD}\mathbf{M}$ are the outputs of the HHD, ${}^{FT}\mathbf{R}_{HHD}$ is the rotational matrix between LRS_{HHD} and LRS_{FT} , and ${}^{FT}\mathbf{o}_{HHD}$ is the origin of LRS_{HHD} expressed in LRS_{FT} .

From ${}^{FT}\mathbf{F}$ and ${}^{FT}\mathbf{M}$, the following indices were defined.

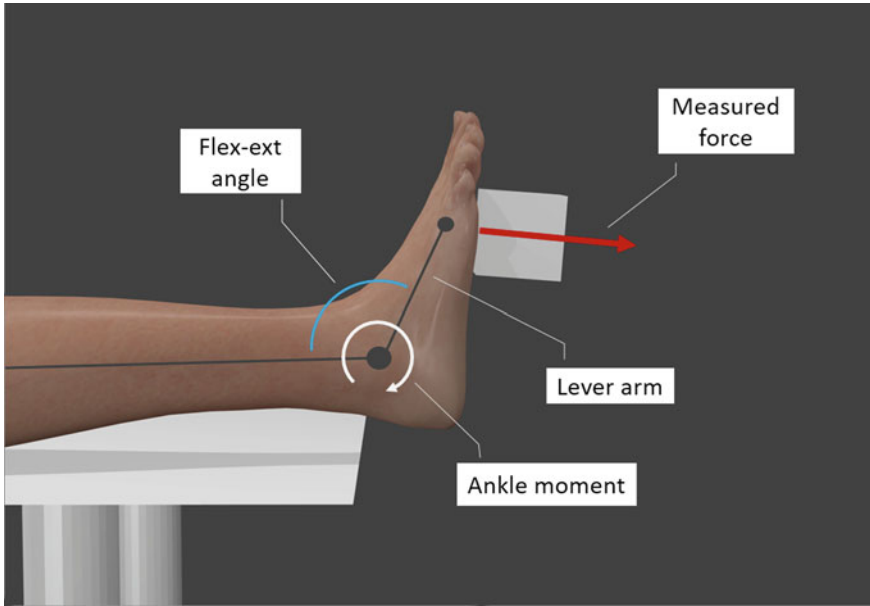


Fig. 2.36 3D rendering representing the parameters computed in ankle strength assessment trials (Ancillao et al. 2017a)

- F_M , as the maximum value of ${}^{\text{FT}}F_y$, which represents the measure of the ankle strength
- F_T , the transverse component of the force exerted by the subject. It represents the intensity of lateral forces that cannot be gathered by means of a single-component load cell:

$$F_T = \sqrt{{}^{\text{FT}}F_x^2 + {}^{\text{FT}}F_z^2} \quad (2.16)$$

- M_M , as the maximum value of ${}^{\text{FT}}M_z$, which represents the ankle dorsiplantar flexion moment when the strength measure is performed
- M_T , the transverse component of the ankle moment:

$$M_T = \sqrt{{}^{\text{FT}}M_x^2 + {}^{\text{FT}}M_y^2} \quad (2.17)$$

All these parameters were averaged across the five repetitions for each subject. As for the kinematic parameters, we computed the coefficient of variation for all the kinetic parameters, to assess the repeatability of the procedure.

As for the knee strength, in order to simulate the strength measurements that are usually gathered in clinical routine, the output of a uniaxial HHD was simulated by considering only the force measured on the z -axis of the HHD and ignoring the other force components and the moments measured by the load cell. The maximum value of force was assumed as the nominal strength measurement ($\hat{F}$), that is, the one considered in clinical routine (Eq. (2.18)). The respective nominal knee moment ($\hat{M}$) was estimated by multiplying $\hat{F}$ by the lever arm (d) between the centre of the HHD and the ankle joint; d was measured with a tape measure as done in clinical routine (Eq. (2.19)).

$$\hat{F} = {}^{\text{HHD}}F_z \quad (2.18)$$

$$\hat{M} = d \cdot \hat{F} \quad (2.19)$$

The differences between the nominal $\hat{F}$, $\hat{M}$, and the respective reference values obtained using the proposed validation procedure (F_M and M_M), were quantified in terms of the root mean square error:

$$\begin{aligned} \text{RMSE}_F &= \sqrt{\frac{\sum_{i=1}^N (F_M^i - \hat{F}_i)^2}{N}} \cdot \frac{100}{\max_i(\hat{F}_i)} (\%) \\ \text{RMSE}_M &= \sqrt{\frac{\sum_{i=1}^N (M_M^i - \hat{M}_i)^2}{N}} \cdot \frac{100}{\max_i(\hat{M}_i)} (\%) \end{aligned} \quad (2.20)$$

Finally, a quality index, namely Q_{index} was defined also for ankle strength measurements, in order to provide an overall quantification of the quality of the strength measurements. Specifically, Q_{index} for ankle strength measurements (Eq. (2.21)) takes into account both the angular displacements of the HHD and the transverse component of moment. The higher the Q_{index} , the higher is the quality of the strength measurement. Its ideal value is 100%.

$$Q_{\text{index}} = 100 \left(1 - \sqrt{\left(\frac{\delta A_1}{90} \right)^2 + \left(\frac{\delta A_2}{90} \right)^2 + \left(\frac{M_T}{M_M} \right)^2} \right) \quad (2.21)$$

2.4.3 Statistics

Descriptive statistics were computed for each index among the subjects. All data were tested for normality by the Shapiro–Wilk test. The significance level was set at 0.05 for all statistical tests. The paired t -test was then computed to check differences of all parameters between the knee extension and the knee flexion trials.

To study the influence of incorrect positioning of the HHD on the measurements of strength and knee moment, we computed the Pearson product–moment correlation coefficient between the kinematic indices (δA_1 , δA_2 , and RoM) and the indices directly related to the inaccuracy of the strength measurement (RMSE_F, RMSE_M, F_T , and M_T). The coefficient r ranges from -1 to 1 (values close to 1 or -1 represent a strong correlation between the variables). The following categorisation for the Pearson coefficient r was considered, as suggested in the literature (Dancey and Reidy 2004): $|r| = 1$: perfect; $0.7 \leq |r| \leq 0.9$: strong; $0.4 \leq |r| \leq 0.6$: moderate; $0.1 \leq |r| \leq 0.3$: weak; $|r| = 0$: zero.

2.4.4 Software Implementation

The workflow for processing strength trials was implemented entirely in MATLAB (The MathWorks, USA).

Processing strength trials requires processing a static (calibration) trial for the subject and the dynamic (measure of strength) trials. Each trial, after recording by the Vicon System, was preprocessed (noise filtering, labelling, etc.) by the Vicon Nexus software and then stored within a C3D file, that is, a file format serving as a container for motion capture data. C3D files for static trial and dynamic trials were imported into MATLAB by means of some ad hoc designed libraries and then processed according to the numerical algorithms previously described.

Software was developed according to a modular criterion: a ‘main’ script accepts filenames and identifies calibration trial and dynamic trials and then uses C3D accessing libraries (implemented as an external function) to extract relevant tracks

from containers and store them in a memory structure. After the details of the trial are identified (left or right side, extension, flexion, etc.) data are passed to another external script that does the processing according to the specific kind of trial. Results of the processing are stored within another memory structure and then passed to other modules (MATLAB functions) that are designed to: (i) format and export processed data to CSV and Excel format, (ii) visualise data through plots, (iii) arrange a visual report that summarises both the numerical and graphical results, and (iv) do further processing on the data, such as descriptive statistics.

The processing software was arranged in the form of a graphical user interface (GUI; Fig. 2.37) in order to speed up multiple trial loading, selection of the kind of trial, and export of the processed results. The GUI is meant to be used by clinicians for routine strength measurements in order to achieve fast and reliable data processing as well as a fast export of results in the form of a clinical report (Fig. 2.38).

Three-dimensional data visualisation was obtained by integrating into the GUI the Open Source code ‘Mokka’ (Barre and Armand 2014) that allows an accurate viewing of 3D tracks and simultaneous visualisation of analog data (Fig. 2.39).

An issue that was encountered in data processing was the identification of the ‘trial start’ and ‘trial end’ events within the recording. In fact, in order to process the force tracks correctly, it was necessary to identify the starting time of the force profile (rising from zero), the ending time (approaching zero), and time at the peak

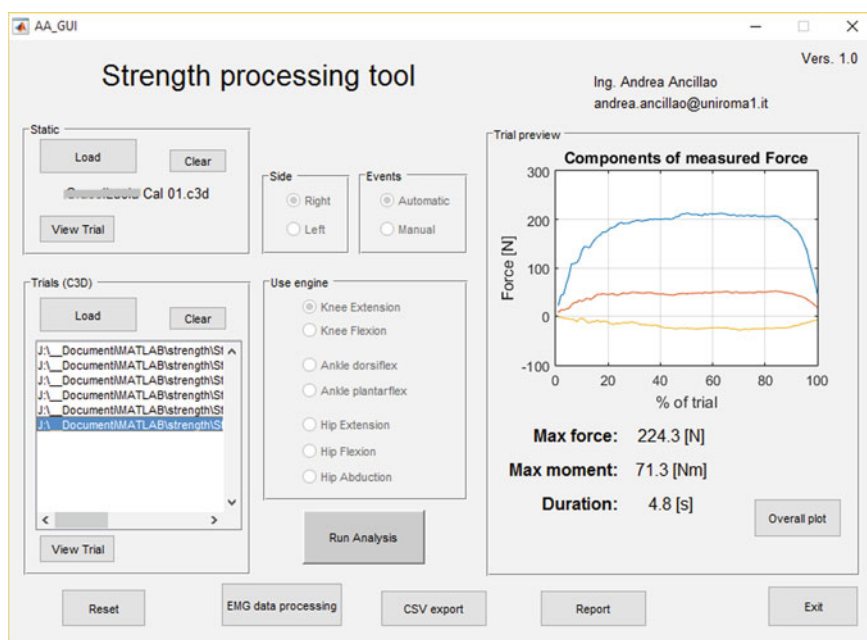


Fig. 2.37 GUI designed for strength data processing, export, and visualisation. Calibration (static) and six dynamic trials were processed and results were displayed

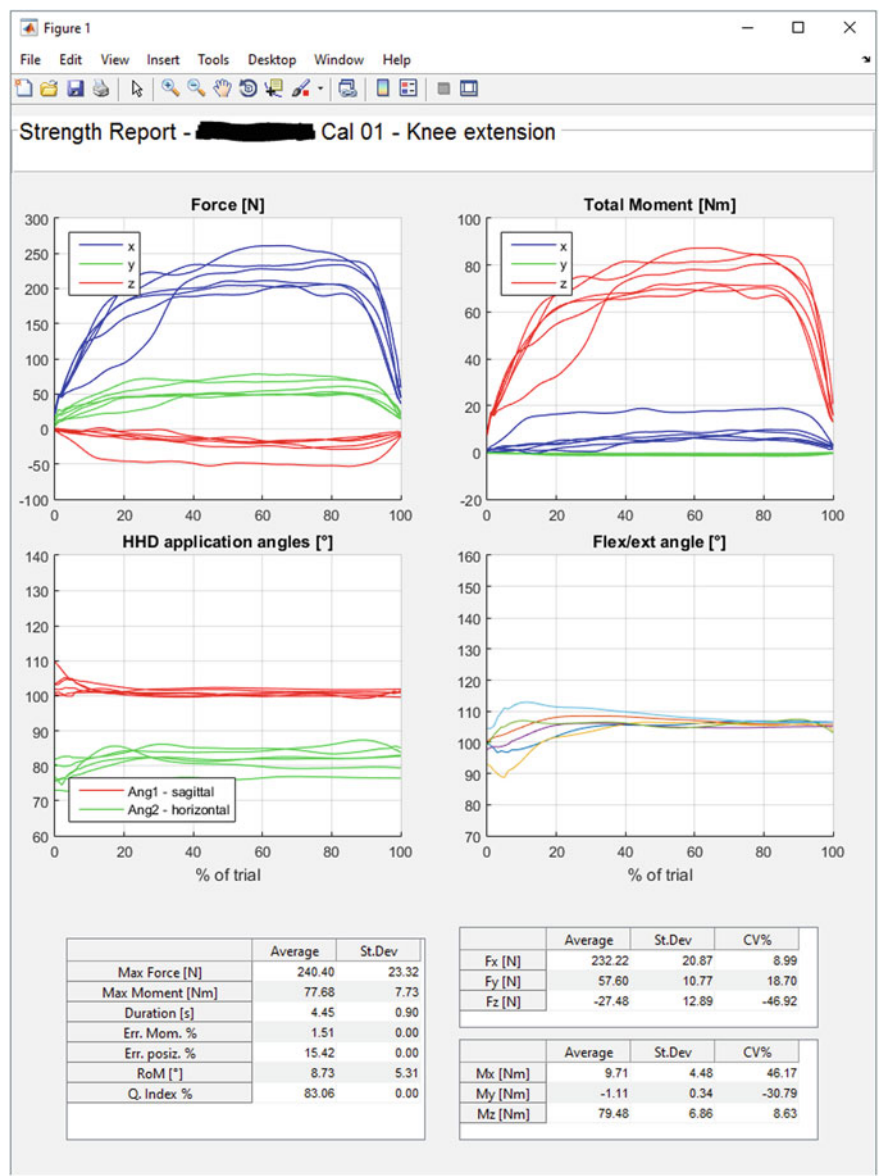


Fig. 2.38 Example of a clinical report. Results from repeated measurements are visualised as well as average and peak values of the parameters

value. Identification was made difficult by the noise superimposed on the analog tracks. As a first step, this issue was solved by asking the user to identify the events manually. The force profile and the distance between the HHD and the limb were shown onscreen and the user had to click on the starting point (force rising and

distance approaching minimum) and ending point (force approaching zero and distance increasing).

This method was proved to work fine, but it significantly slowed data processing. In order to shorten data processing time and the need of user intervention, an algorithm for the automatic identification of events was developed. The algorithm required: (i) noise filtering of the force track, (ii) automatic removal of artefacts by means of thresholds, (iii) computing of the first-order derivative of the force profile, and (iv) identification of the rising and decreasing parts of the force profile. This algorithm worked on most trials, but sometimes it failed. The most adverse conditions were observed for the tracks with a low signal-to-noise ratio or tracks with strong artefacts on the force profile, such as the accidental application of force on the sensible surface of the load cell before starting the measurement. For this reason, the possibility of manually identifying events was kept in the final implementation of the GUI (Fig. 2.37).

In order to explain visually the measurement setup in published papers, presentations, and also this book, several fine artworks were created. High-quality three-dimensional representations of human characters were obtained by means of ‘Make Human’ Open Source graphic software (<http://www.makehuman.org/>), in the form of high-resolution meshes (.stl, .mnh, .dae formats).

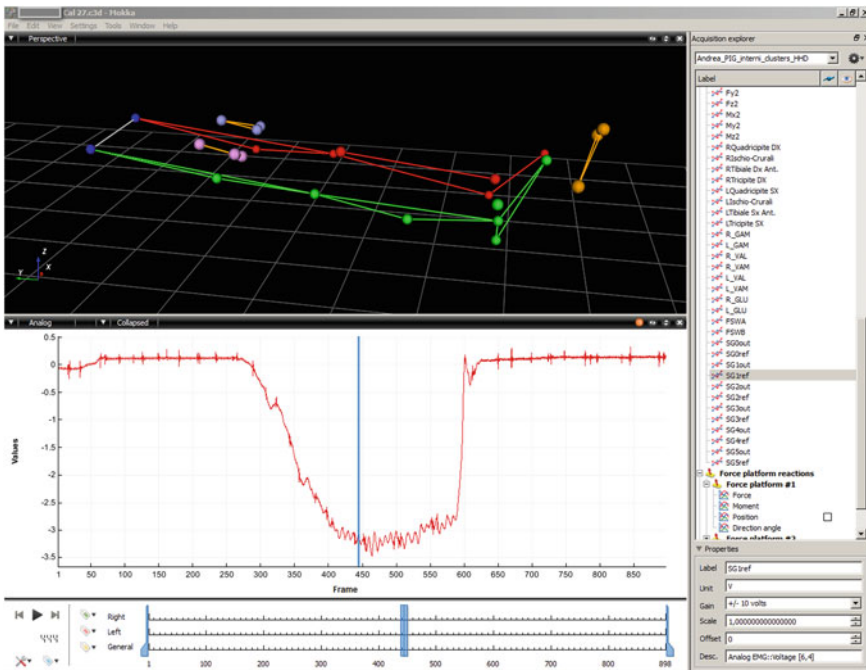


Fig. 2.39 3D visualisation of an ankle plantarflexion strength trial. The RAW force profile is shown at the bottom

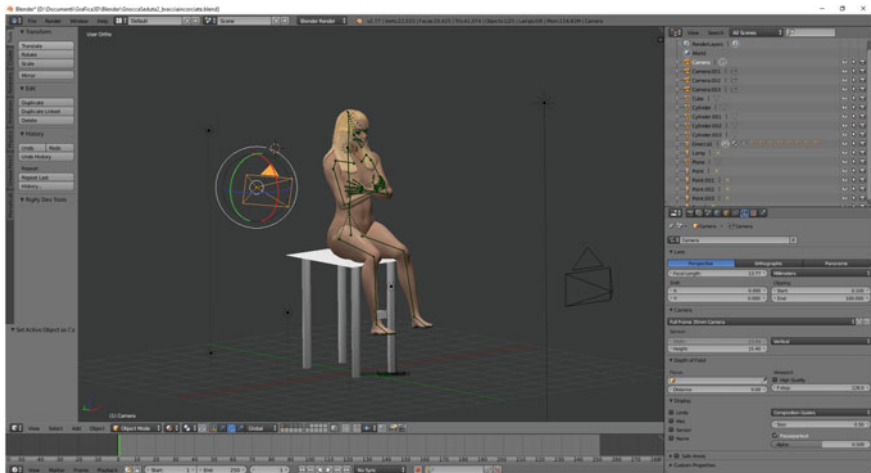


Fig. 2.40 Screenshot of Blender software illustrating the graphical design, workflow, and rendering of human characters performing a knee strength measurement

Characters were posed, dressed, and rendered by using ‘Blender’ Open Source 3D CAD software (www.blender.org). Examples of character posing and scene creation for illustrating the knee strength measurements are shown in Fig. 2.40.

2.5 Results and Discussion

2.5.1 Knee Strength

Results of the quality analysis of knee strength trials are reported in Table 2.4. Displayed values are average and SD among the subjects included in the study; p values of the t -test between movements are also reported. The Shapiro–Wilk test showed that kinetic and kinematic indices had a normal distribution across the subjects, therefore parametric statistics could be applied.

From the kinematic analysis (Table 2.4) it can be observed that the RoM, that is, the angular variation of the knee flexion/extension angle across the trial, was never close to 0° and it was higher for the extension movement with respect to the flexion one and with a statistically significant difference. Moreover, the standard deviation of RoM was higher for knee extension.

The positioning error index δA_1 showed comparable values between knee flexion and extension. Instead, the δA_2 index came out higher for knee flexion. Values of δA_2 were always higher than δA_1 for both flexion and extension.

No differences between flexion and extension were observed for the CV_{RoM} but, as a trend, it had higher values than the respective coefficients of variation obtained

Table 2.4 Kinematic and kinetik indices

	Extension	Flexion	<i>t</i> -test
RoM (°)	21.7 (9.8)	15.5 (6.4)	0.005*
δA_1 (°)	5.7 (3.4)	6.1 (3.8)	0.712
δA_2 (°)	9.3 (6.0)	15.1 (8.3)	0.014*
CV_{RoM} (%)	23.2 (12.4)	23.5 (11.0)	0.919
$CV_{\delta A_1}$ (%)	5.0 (3.7)	2.1 (1.2)	0.002*
$CV_{\delta A_2}$ (%)	4.8 (2.4)	3.3 (1.3)	0.013*
$\hat{F}$ (N)	249.4 (27.3)	146.4 (23.9)	$\ll 0.001^*$
$\hat{M}$ (Nm)	88.4 (12.4)	49.9 (8.3)	$\ll 0.001^*$
F_M (N)	242.5 (28.8)	145.1 (26.4)	$\ll 0.001^*$
F_T (N)	63.0 (22.8)	37.5 (14.1)	$\ll 0.001^*$
M_M (Nm)	87.8 (12.5)	48.7 (9.5)	$\ll 0.001^*$
M_T (Nm)	14.6 (8.4)	9.4 (4.7)	0.018*
RMSE _F (%)	3.0 (2.6)	3.3 (1.7)	0.613
RMSE _M (%)	4.0 (1.7)	5.2 (2.4)	0.035*
$CV_{\hat{F}}$ (%)	8.1 (6.0)	5.9 (3.2)	0.092
$CV_{\hat{M}}$ (%)	8.2 (5.9)	5.8 (3.0)	0.068
Q_{index} (%)	82.8 (11.2)	75.3 (14.1)	0.069

Mean (SD) values for the kinematic and kinetic indices measured for the knee extension and flexion. RMSE values were evaluated between one-component and six-component measurements

*Indicates a statistically significant difference between extension and flexion ($p < 0.05$). The p values are reported in the t -test column

for the positioning errors, that is, $CV_{\delta A_1}$ and $CV_{\delta A_2}$. A statistically significant difference was observed between flexion and extension for both $CV_{\delta A_1}$ and $CV_{\delta A_2}$.

The kinetic analysis (Table 2.4) showed that force and moment parameters $\hat{F}$, $\hat{M}$, F_M , F_T , M_M , and M_T , were significantly higher for knee extension than the knee flexion. All RMSEs values were less than 5% (except for the RMSE_M of knee flexion which was slightly higher). A statistically significant difference was observed for the RMSE_M of knee flexion that was higher than knee extension. RMSE_F were lower than RMSE_M and no significant differences were observed between the two rotations. Considering the coefficients of variation, $CV_{\hat{F}}$ and $CV_{\hat{M}}$ were always lower than 10%. Both CVs were higher for knee extension; however, no significant differences were observed. The average Q_{index} was relatively high for both extension and flexion even though it was slightly lower for knee flexion.

Correlation analysis was run between the misplacement parameters (δA_1 and δA_2) and RoM, and the ones directly related to the inaccuracy of the strength measurement (RMSE_F, RMSE_M, F_T , and M_T). Results are reported in Table 2.5. No correlation was observed between RoM and the kinetic parameters for both extension and flexion. For the knee extension, a strong correlation was observed between δA_2 and F_T , M_T , RMSE_F parameters. Correlation was also strong between

Table 2.5 Correlation coefficients

		F_T	M_T	$RMSE_F$	$RMSE_M$
Knee Ext	RoM	-0.1	0.0	-0.1	0.1
	δA_1	0.8**	0.5*	0.7**	0.1
	δA_2	0.7**	0.8**	0.9**	0.3
Knee Flex	RoM	0.0	0.0	0.0	-0.1
	δA_1	0.2	0.0	0.3	-0.1
	δA_2	0.5*	0.8**	0.5*	0.4*

Pearson correlation coefficients (r) between kinematic indices and kinetic indices for knee extension and knee flexion

*Indicates a moderate correlation ($0.4 \leq |r| \leq 0.6$)

**Indicates a strong correlation ($0.7 \leq |r| \leq 0.9$)

δA_1 and F_T , $RMSE_F$ although it was moderate between δA_1 and M_T . Low correlations with $RMSE_M$ were observed for each misplacement parameter. As concerns the knee flexion, significant correlations were observed only for the δA_2 . Specifically, the correlation was strong with M_T whereas it was moderate with the other parameters.

2.5.2 Ankle Strength

The Shapiro–Wilk test run on ankle data across subjects proved it to be normally distributed; therefore t -tests could be applied. Average and standard deviations across subjects of both kinematic and kinetic parameters, including the p -values of differences, are reported in Table 2.6.

As shown in Table 2.6, the ankle did not remain still and moved during the measurements as the observed RoMs were not 0° . This meant that the operator was not able to keep the HHD and the foot completely still, leading to an undesired motion of the foot during the trial. This finding was in line with the results of Kim et al. (2014) which demonstrated a decreased measurement validity when the dynamometer was not fixed but held in-hand by the operator.

The observed RoM was slightly higher for plantarflexion trials ($p = 0.06$), where a higher exerted force was registered ($p < 0.01$).

Angular displacements δA_1 and δA_2 were higher in plantarflexion trials than dorsiflexion ones (Table 2.6). This was attributed to the higher force exerted in plantarflexion trials that reduced the operator's ability to keep the HHD in place during the measurements. On the contrary, the operator had more control over the dorsiflexion trials because angular displacements were lower.

As regards $\hat{F}$ and $\hat{M}$, they were higher than F_M and M_M for both directions, and the transversal components F_T and M_T were not negligible. $\hat{F}$ and $\hat{M}$ represented the force and moment that are commonly measured in a strength assessment by means of clinical HHDs. In case of misplacement, $\hat{F}$ and $\hat{M}$ may differ significantly from the force and moment effectively exerted by the joint. The findings of this study showed that a wrong positioning of the HHD increased the lateral

Table 2.6 Kinematic and kinetic indices

	Plantar-flexion	Dorsiflexion	<i>t</i> -test
RoM (°)	26.7 (9.9)	21.1 (6.1)	0.06
δA_1 (°)	29.5 (8.7)	5.1 (2.9)	<0.01*
δA_2 (°)	12.9 (5.4)	5.1 (3.3)	<0.01*
CV_{RoM} (%)	17.6 (10.0)	21.1 (9.9)	0.31
$CV_{\delta A_1}$ (%)	16.2 (10.0)	50.8 (29.1)	<0.01*
$CV_{\delta A_2}$ (%)	26.1 (19.1)	49.1 (25.2)	<0.01*
$\hat{F}$ (N)	244.3 (46.2)	191.8 (38.5)	<0.01*
$\hat{M}$ (Nm)	54.0 (15.7)	36.5 (10.1)	<0.01*
F_M (N)	209.7 (39.4)	189.8 (39.2)	0.14
F_T (N)	125.0 (31.2)	34.3 (13.9)	<0.01*
M_M (Nm)	30.5 (4.8)	23.7 (4.9)	<0.01*
M_T (Nm)	12.9 (4.4)	3.0 (1.3)	<0.01*
RMSE _F (%)	13.3 (5.4)	1.6 (1.3)	<0.01*
RMSE _M (%)	35.3 (11.1)	29.4 (11.8)	0.14
$CV_{\hat{F}}$ (%)	7.2 (5.1)	8.1 (4.4)	0.57
$CV_{\hat{M}}$ (%)	16.7 (12.1)	15.6 (7.1)	0.75
Q_{index} (%)	44.2 (11.0)	88.2 (5.0)	<0.01*

Mean (SD) values for parameters measured for the ankle plantarflexion and dorsiflexion. The *p* values are reported in the last column. Significant differences (*p* < 0.05) are starred

components reducing the force on the principal axis. The highest differences between $\hat{F}$ and F_M and between $\hat{M}$ and M_M were observed in the case of plantarflexion trials. Lower differences were observed in the case of dorsiflexion. This finding was confirmed by the magnitudes of lateral components of force and moment F_T and M_T that were higher for plantarflexion than dorsiflexion. These findings implied that some relevant issues occurred during these trials. Such issues were identified as the wrong angular positioning on both planes, represented by the δA_1 and δA_2 coefficients that were both high in plantarflexion (Table 2.6). These results suggested a higher reliability of ankle dorsiflexion trials than the plantarflexion ones.

As regards the accuracy of the measurements, which is represented by the RMSE values, a very low value of RMSE_F in dorsiflexion (<5%) was observed, whereas it was higher for plantarflexion (<15%) confirming that the ankle strength assessment was more accurate for dorsiflexion than for plantarflexion. This was attributed to the higher magnitude of lateral components of force exerted by the ankle in the latter movement.

Variability within the same subject was quantified by means of CV coefficients. The highest values of CVs were observed for $CV_{\delta A_1}$ and $CV_{\delta A_2}$ during dorsiflexion trials (~50%). This result proved to have poor repeatability in terms of HHD positioning also when the operator was able to maintain the HHD motionless, demonstrating that the strength measurements are likely influenced by the strength

and experience of the examiner, in accordance with the findings of other studies (Hébert et al. 2011; Marmon et al. 2013). Average values of $CV_{\hat{F}}$ were less than 10% and average values of $CV_{\hat{M}}$ were less than 20% for both plantarflexion and dorsiflexion, indicating a good intrasubject repeatability of force measurement. The repeatability coefficients of moments were also good but lower than forces, likely due to the wrong positioning of the HDD.

Regarding the knee strength trials, the quality coefficient Q_{index} was computed. It is a synthetic index that represents the overall quality of the measurement. Its values are reported in Table 2.6. It was conceived to take into account both the angular misplacements of the HDD and the undesired lateral components of moment. Its average value resulted lower for plantarflexion than dorsiflexion. These results were in accordance with the other parameters that identified the most relevant inaccuracies in the ankle plantarflexion trials. Again, this finding was in agreement with the findings of other works that reported poor repeatability and reliability of ankle strength measurements, especially for plantarflexion trials (Hébert et al. 2011; Marmon et al. 2013).

Correlation coefficients were computed in order to check the linear dependence between kinetic and kinematic parameters. Results of correlation analysis are reported in Table 2.7. A strong correlation was found between the RoM and $RMSE_M$ indicating that the intensity of the undesired motion of the foot had effect on the measured moment. It was assumed as due to the variation that occurred in the lever arm and wrong positioning of the sensor on the sagittal and horizontal planes.

2.5.3 Influence of the Operator’s Ability on Strength Measurements

In order to test the operator-dependent inaccuracies, when performing a strength measurement as a ‘make test’ method, the RoM of the limb was measured as the subject’s limb should remain motionless across the trial. Ideally, the RoM should be

Table 2.7 Correlation table

		F_T	M_T	$RMSE_F$	$RMSE_M$
Ankle Plantarflexion	RoM	0.1	0.3	0.0	−0.3
	δA_1	−0.1	−0.5	0.3	−0.1
	δA_2	0.3	0.3	0.1	0.0
Ankle Dorsiflexion	RoM	−0.2	−0.1	−0.3	0.7**
	δA_1	0.1	−0.4*	0.3	0.0
	δA_2	0.4*	0.4*	0.4*	−0.3

Pearson correlation coefficients (r) between kinematic indices and kinetic indices for the ankle plantarflexion and dorsiflexion

*Indicates a moderate correlation ($0.4 \leq |r| \leq 0.6$)

**Indicates a strong correlation ($0.7 \leq |r| \leq 0.9$)

close to 0° . Knee RoM was statistically higher for extension trials, in which the force exerted was higher, than flexion ones (Table 2.5), indicating that the operator was not able to counteract completely the participant's force and therefore he was not able to hold the limb completely still with poor repeatability across the trials. The same issue occurred in ankle strength trials, where RoM was higher in plantarflexion trials.

Regarding the direct measurement of HHD placement, the positioning error δA_1 showed comparable values between knee flexion and extension, implying the same difficulty level for the operator in correctly positioning the HHD on the sagittal plane during the two types of trial. The variability of the index was relatively low, indicating a good repeatability in the angular positioning of the HHD on the sagittal plane. δA_2 was found higher than δA_1 and was statistically higher for knee flexion, indicating that the most relevant source of inaccuracy was the HHD misplacement on the horizontal plane especially in the case of knee flexion. This may be due to the uncomfortable position that the operator had to assume to hold the HHD behind the subject's ankle in knee flexion trials. It follows that the operator has to pay special attention to avoid the rotation of the HHD on the horizontal plane. In ankle strength trials, both angular displacements δA_1 and δA_2 were higher in plantarflexion trials, corresponding to higher exerted force and higher motion of the limb. Again these coefficients demonstrated the operator's difficulties in counteracting the patient's force. Instead, the operator had more control on the dorsiflexion trials where the force was lower and smaller displacements were observed.

Considering the coefficients of variation CV of the outputs of the knee strength measurements (CV_F and CV_M), a low level of variability was observed ($\leq 5\%$) showing a good intrasubject repeatability of measurements for both knee flexion and knee extension trials, in agreement with the literature outcomes (Kim et al. 2014; Martin et al. 2006; Phillips et al. 2000). Moreover, CV_F and CV_M of knee flexion were lower than knee extension, indicating that knee extension trials had more inherent critical issues with respect to knee flexion ones. This finding can be interpreted by observing that the force $\hat{F}$ and the moment $\hat{M}$ values exerted by the subjects involved in the present study were higher during the extension trial than the flexion one implying a greater difficulty for the operator to maintain the participant's limb motionless. This finding is in agreement with the results of Laing et al. (1995) that showed higher quality of the trials achieved by fixing the HHD in contrast to the HHD freely held by the operator. In ankle strength, the highest values of CVs were observed for $CV_{\delta A_1}$ and $CV_{\delta A_2}$ during dorsiflexion trials ($\sim 50\%$). This result confirmed a poor repeatability in HHD positioning, demonstrating that the strength measurements are likely influenced by the strength and experience of the examiner, in accordance with the findings of other studies (Hébert et al. 2011; Marmon et al. 2013).

2.5.4 Use of a One-Component HHD Versus Six-Component Load Cell

The inaccuracy associated with a uniaxial HHD in comparison with the values collected via a six-component HHD could be quantified by F_T , M_T , $RMSE_F$, and $RMSE_M$. The first two indices represented the lateral components of the force and moment that are commonly neglected when a uniaxial HHD is used. $RMSE_F$ and $RMSE_M$, instead, allowed the quantification of the accuracy of strength and moment measurements performed with the uniaxial HHD in the clinical routine comparing them with the actual ones obtained by a six-component HHD and the OS.

Focusing on F_T and M_T values of knee strength analysis, the highest values were obtained during the extension trials confirming the above reported findings on the higher complexity of extension trials. $RMSE_F$ and $RMSE_M$ were relatively low, always $\leq 5.2\%$, for both knee flexion and knee extension. We can therefore speculate that the uniaxial HHD is reliable and accurate enough for use in clinical contexts, according to the dataset acquired in the present study.

In the case of ankle strength, the highest values of F_T and M_T were observed during the plantarflexion trials, confirming it as the worst case. This finding was confirmed by the low value of $RMSE_F$ observed in ankle dorsiflexion ($<5\%$), and the same parameter was higher for plantarflexion ($<15\%$) showing that the ankle strength assessment by means of a single-component HHD was more accurate for dorsiflexion than for plantarflexion. The inaccuracy observed for plantarflexion trials is not negligible, therefore the use of a single-component HHD is not recommended for ankle plantarflexion strength assessment. This result is in agreement with other studies on ankle strength reliability conducted by other methods (Hébert et al. 2011; Marmon et al. 2013).

In order synthetically to describe the quality of a strength measurement, according to the previously discussed parameters, we computed a synthetic quality index Q_{index} that takes into account the undesired angular displacements and the undesired lateral components of moment. The average Q_{index} value was relatively high for both knee extension and knee flexion, without statistical difference. This finding supported the conclusions that the inaccuracies due to both the positioning of the HHD and the lateral force and moment components can be considered negligible. Instead, the Q_{index} was lower for ankle trials and very low for ankle plantarflexion ($Q_{\text{index}} = 44.2$) confirming that this is the most critical condition and the HHD method is not recommended for ankle strength assessment.

2.5.5 Correlation Between Improper Positioning and Strength Measurement

Correlation between kinematic and kinetic parameters was analysed to investigate the influence of the HHD misplacement and the accuracy of the uniaxial HHD in the strength measurements.

For the knee extension, a strong correlation was found between δA_2 , δA_1 , and F_T , M_T , $RMSE_F$. The correlation was low with $RMSE_M$ (Table 2.5). These results show that the angular misplacements had effect on the lateral undesired components of force and moment, whereas the error on the actual moment was not affected by the incorrect orientation of the HHD. The misplacements affected the force measurement to a greater extent than the moment one. Moreover, the RoM had no influence on the lateral components of force/moment or effect on the RMSEs. This means that the range of motion, if it is maintained within the values observed in this work, does not affect the measurements in terms of lost information due to lateral components, which are not measured by the commercial HHDs used in clinical practice. In knee flexion trials, correlations were observed between δA_2 and all the parameters. The correlation was strong only versus M_T , and it was moderate towards the other parameters. The strong correlation between δA_2 and M_T , that was observed also for the extension trial, demonstrated that the main misplacement of the HHD is on the horizontal plane and an increase of δA_2 could affect the quality of strength measurements performed with a uniaxial HHD. In fact, it had effect mainly on the lateral components of moment and it is therefore a critical positioning parameter to consider while gathering data. Conversely, the absence of correlation for δA_1 and RoM may be connected to the lower forces and moments exerted in the case of knee flexion. As in the knee extension, the RoM of knee flexion did not affect the lateral components of force and moment if it is maintained within the values found in the present study. The HHD misplacement quantified via the δA_2 index appears to be the main critical parameter for the quality assessment of a knee strength measurement.

For ankle plantarflexion, no significant correlation was observed for the parameters, whereas a strong correlation was found between the RoM and $RMSE_M$ in ankle dorsiflexion (Table 2.7). The absence of correlation for dorsiflexion trials could be attributed to the low quality, low repeatability, and issues observed in those trials. In fact the operator was not able to counteract the patient force correctly, therefore misplacements were the largest. As in knee flexion, moderate correlations were observed for the δA_2 index of ankle dorsiflexion trials, confirming this as the most critical parameter for the quality of strength trials. The high correlation between RoM and $RMSE_M$ was assumed as due to the variation that occurred in the lever arm and wrong positioning of the sensor on sagittal and horizontal planes.

2.6 Conclusion

This work was aimed at investigating the measurement issues occurring in knee and ankle strength measurements. A motion capture protocol involving an optoelectronic system and a multicomponent handheld load cell was developed and validated in order to assess the sources of inaccuracy that occur when a clinical one-component HHD is used to measure strength. The protocol allowed the direct

measurement of displacement, wrong positioning of the HHD, motion of the subject during the trial, and the magnitude of the lateral components of force and moment that are commonly neglected in clinical measurements. The concurrent measure of displacements and kinetic quantities does not interfere with the strength measurement, therefore the method may be used in the clinical routine to ensure the quality of each recorded trial. As a ‘quality index’ is assigned to each trial this makes it easy for the clinician to identify and discard the unreliable trials, improving the overall quality of the final measurements. This method was validated on knee and ankle strength measurements conducted on adult healthy subjects who represent the most aversive case for the operator due to the relatively high force they can exert (Ancillao et al. 2014, 2015, 2017a, b).

This study showed that the limbs of subjects did not remain perfectly still during the measurements and this represented a nonnegligible source of inaccuracy. According to the data collected in this study, it can be concluded that the use of uniaxial HHDs can be assumed reliable and accurate enough for both knee flexion and extension, if lower limb displacements and HHD misplacement are kept within the values found in the present study (see Table 2.4). The use of a uniaxial HHD for the strength measurement, in place of a six-component one, can be considered a reliable method when a maximum value of inaccuracy equal to 6% is considered acceptable. The most critical case is the knee extension, where the operator is required to pay strong attention in positioning the HHD on the horizontal plane. In conclusion, even though the operator was not able to keep the limb of the subject perfectly still and the HHD actual orientation was different from the desired one, the measurement outputs were reliable and accurate enough for both knee flexion and extension.

Different results were observed for ankle strength measurements, where the highest sources of error were identified in the angular displacements of the HHD and in the undesired motion of the foot. Worst cases were observed in plantarflexion trials. Thus, the quality of ankle strength measurements was strongly dependent on the operator, and his ability to hold the HHD in place. Based on the current results, ankle dorsiflexion trials could be considered more reliable than the plantarflexion ones. Ankle plantarflexion came up with a lower quality index to which corresponded higher RMSE values and higher intensity of lateral components of force and moment. The inherent reliability of HHDs used for the measurement of plantarflexion strength is consequently low. Thus, commercial one-component HHDs should be used carefully when chosen as the preferred measurement method to estimate ankle strength. The clinical protocol should be carefully revised in order to ensure proper limb fixation and to reduce both the effects of foot motion and the HHD positioning errors. Ankle dorsiflexion strength assessment by means of one-component HHD could be considered reliable, as F_T , M_T , and $RMSE_F$ measured in this study were relatively low. In confirmation, differences between the $\hat{F}$ and the F_M were low and the average quality index was relatively high. Thus, the error could be considered acceptable for the clinical use of the one-component HHDs. However, it is always recommended to pay attention in

HHD positioning, especially on the angular positioning on both horizontal and sagittal planes.

The main limitations of this study were: the absence of a gold-standard reference, such as the isokinetic dynamometer, to compare the values of force measured by the HHD, and the absence of an interoperator repeatability study. Further study may therefore involve the analysis of interoperator repeatability by comparing the kinematic and kinetic parameters obtained by different expert operators. Moreover, this study was conducted on healthy adult subjects but some other positioning difficulties and inaccuracies may arise in the case of pediatric subjects or subjects with pathology.

The method and the results discussed in this work may lead to a better understanding of HHD measurements and provide directions to clinicians for the proper use of the instrument. Further steps may involve analysis of inaccuracies associated with different anatomical districts and the quality analysis of strength measurements conducted on patients with pathology.

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References

- Allen G, Gandevia S, McKenzie D (1995) Reliability of measurements of muscle strength and voluntary activation using twitch interpolation. *Muscle Nerve* 18:593–600
- Amundsen LR, Patterson RP, Baxter TL, Scudder GN, Duescher WO, Dahlman WE, Schukar GW, Steinback CI (1987) Method and apparatus for measuring the isometric muscle strength of multiple muscle groups in the human body. Patent US4702108. Inventor: Amundsen et al. 4702108
- Ancillao A, Galli M, Celletti C, Castori M, Albertini G, Camerota F (2012) Temporomandibular joint mobility in adult females with Ehlers-Danlos syndrome, hypermobility type (also known as joint hypermobility syndrome). *J Cranio-Maxillary Dis* 1:88–95
- Ancillao A, Galli M, Vimercati SL, Albertini G (2013) An optoelectronic based approach for handwriting capture. *Comput Methods Programs Biomed* 111:357–365
- Ancillao A, Patanè F, Rossi S, Pacilli A, Cappa P (2014) Lower limb strength measurements by Hand Held Dynamometer assisted by optoelectronic system. In: *MMT2014*, pp. 1–2
- Ancillao A, Rossi S, Patanè F, Cappa P (2015) A preliminary study on quality of knee strength measurements by means of Hand Held Dynamometer and Optoelectronic System. In: *IEEE—MeMeA2015*, pp. 595–599
- Ancillao A, Palermo E, Rossi S (2017a) Validation of ankle strength measurements by means of a hand-held dynamometer in adult healthy subjects. *J Sens* 2017:1–8
- Ancillao A, Rossi S, Cappa P (2017b) Analysis of knee strength measurements performed by a hand-held multicomponent dynamometer and optoelectronic system. *IEEE Trans Instrum Meas* 66:85–92

- Baltzopoulos V, Brodie DA (1989) Isokinetic dynamometry. Applications and limitations. *Sport Med* 8:101–116
- Bandinelli S, Benvenuti E, Del Lungo I, Baccini M, Benvenuti F, Di Iorio A, Ferrucci L (1999) Measuring muscular strength of the lower limbs by hand-held dynamometer: a standard protocol. *Aging Clin Exp Res* 11:287–293
- Barre A, Armand S (2014) Biomechanical ToolKit: open-source framework to visualize and process biomechanical data. *Comput Methods Programs Biomed* 114:80–87
- Bohannon RW (1986) Test-retest reliability of hand-held dynamometry during a single session of strength assessment. *Phys Ther* 66:206–209
- Bohannon RW (1988) Make tests and break tests of elbow flexor muscle strength. *Phys Ther* 68:193–194
- Bohannon R (1990) Hand-held compared with isokinetic dynamometry for measurement of static knee extension torque (parallel reliability of dynamometers). *Clin Phys Physiol Meas* 11:217–222
- Bohannon RW, Andrews AW (1987) Interrater reliability of hand-held dynamometry. *Phys Ther* 67:931–933
- Brunner R, Rutz E (2013) Biomechanics and muscle function during gait. *J Child Orthop* 7:367–371
- Camerota F, Galli M, Cimolin V, Celletti C, Ancillao A, Blow D, Albertini G (2015) The effects of neuromuscular taping on gait walking strategy in a patient with joint hypermobility syndrome/Ehlers-Danlos syndrome hypermobility type. *Ther Adv Musculoskelet Dis* 7:3–10
- Capodaglio P, Vismara L, Menegoni F, Baccalaro G, Galli M, Grugni G (2009) Strength characterization of knee flexor and extensor muscles in Prader-Willi and obese patients. *BMC Musculoskelet Disord* 10:47
- Cappozzo A, Cappello A, Della Croce U, Pensalfini F (1997) Surface-marker cluster design criteria for 3-D bone movement reconstruction. *IEEE Trans Biomed Eng* 44:1165–1174
- Clark RA, Bryant AL, Pua Y, McCrory P, Bennell K, Hunt M (2010) Validity and reliability of the Nintendo Wii Balance Board for assessment of standing balance. *Gait Posture* 31: 307–310
- Csuka M, McCarty DJ (1985) Simple method for measurement of lower extremity muscle strength. *Am J Med* 78:77–81
- Dancey C, Reidy J (2004) *Statistics without maths for psychology*. Pearson Education, Harlow
- Davis RB, Öunpuu S, Tyburski D, Gage JR (1991) A gait analysis data collection and reduction technique. *Hum Mov Sci* 10:575–587
- Eek M, Kroksmark A, Beckung E (2006) Isometric muscle torque in children 5 to 15 years of age: normative data. *Arch Phys Med Rehabil* 87:1091–1099
- Fulcher ML, Hanna CM, Raina Elley C (2010) Reliability of handheld dynamometry in assessment of hip strength in adult male football players. *J Sci Med Sport* 13:80–84
- Galli M, Rigoldi C, Brunner R, Virji-Babul N, Giorgio A (2008) Joint stiffness and gait pattern evaluation in children with Down syndrome. *Gait Posture* 28:502–506
- Galli M, Cimolin V, Vismara L, Grugni G, Camerota F, Celletti C, Albertini G, Rigoldi C, Capodaglio P (2011) The effects of muscle hypotonia and weakness on balance: a study on Prader-Willi and Ehlers-Danlos syndrome patients. *Res Dev Disabil* 32:1117–1121
- Hartmann A, Knols R, Murer K, de Bruin ED (2009) Reproducibility of an isokinetic strength-testing protocol of the knee and ankle in older adults. *Gerontology* 55:259–268
- Hébert LJ, Maltais DB, Lepage C, Saulnier J, Crête M, Perron M (2011) Isometric muscle strength in youth assessed by hand-held dynamometry: a feasibility, reliability, and validity study. *Pediatr Phys Ther* 23:289–299
- Hughes VA, Frontera WR, Wood M, Evans WJ, Dallal GE, Roubenoff R, Fiatarone Singh MA (2001) Longitudinal muscle strength changes in older adults: influence of muscle mass, physical activity, and health. *J Gerontol A Biol Sci Med Sci* 56:B209–B217
- Janssen JC, Le-Ngoc L (2009) Intratester reliability and validity of concentric measurements using a new hand-held dynamometer. *Arch Phys Med Rehabil* 90:1541–1547
- Jones CJ, Rikli ER, Beam WC (1999) A 30-s chair-stand test as a measure of lower body strength in community-residing older adults. *Res Q Exerc Sport* 70:113–119

- Kim WK, Kim DK, Seo KM, Kang SH (2014) Reliability and validity of isometric knee extensor strength test with hand-held dynamometer depending on its fixation: a pilot study. *Ann Rehabil Med* 38:84–93
- Laing BA, Mastaglia FL, Lo SK, Zilko P (1995) Comparative assessment of knee strength using hand-held myometry and isometric dynamometry in patients with inflammatory myopathy. *Physiother Theor Pract* 11:151–156
- Mahony K, Hunt A, Daley D, Sims S, Adams R (2009) Inter-tester reliability and precision of manual muscle testing and hand-held dynamometry in lower limb muscles of children with spina bifida. *Phys Occup Ther Pediatr* 29:44–59
- Marmon AR, Pozzi F, Alnahdi AH, Zeni JA (2013) The validity of plantarflexor strength measures obtained through hand-held dynamometry measurements of force. *Int J Sports Phys Ther* 8:820–827
- Martin HJ, Yule V, Syddall HE, Dennison EM, Cooper C, Aihie Sayer A (2006) Is hand-held dynamometry useful for the measurement of quadriceps strength in older people? A comparison with the gold standard Biodex dynamometry. *Gerontology* 52:154–159
- Maughan R, Watson J, Weir J (1983) Strength and cross-sectional area of human skeletal muscle. *J Physiol* 338:37–49
- Mokkink LB, Terwee CB, Knol DL, Stratford PW, Alonso J, Patrick DL, Bouter LM, de Vet HC (2010) The COSMIN checklist for evaluating the methodological quality of studies on measurement properties: a clarification of its content. *BMC Med Res Methodol* 10:22
- Peindl RD, McCarthy ML, MacKenzie JE (1997) Apparatus for exercising and measuring strength of a patient's limb and an adjustable pivot clamp. Patent US5662591. Inventor: Peindl et al. 5662591
- Phillips BA, Lo SK, Mastaglia FL (2000) Muscle force measured using “break” testing with a hand-held myometer in normal subjects aged 20 to 69 years. *Arch Phys Med Rehabil* 81:653–661
- Riddle DL, Finucane SD, Rothstein JM, Walker ML (1989) Intrasection and intersection reliability of hand-held dynamometer measurements taken on brain-damaged patients. *Phys Ther* 69:182–194
- Rigoldi C, Galli M, Cimolin V, Camerota F, Celletti C, Tenore N, Albertini G (2012) Gait strategy in patients with Ehlers-Danlos syndrome hypermobility type and Down syndrome. *Res Dev Disabil* 33:1437–1442
- Spink MJ, Fotoohabadi MR, Wee E, Hill KD, Lord SR, Menz HB (2011) Foot and ankle strength, range of motion, posture, and deformity are associated with balance and functional ability in older adults. *Arch Phys Med Rehabil* 92:68–75
- Terwee CB, Bot SDM, de Boer MR, van der Windt DAWM, Knol DL, Dekker J, Bouter LM, de Vet HCW (2007) Quality criteria were proposed for measurement properties of health status questionnaires. *J Clin Epidemiol* 60:34–42
- Tsaopoulos DE, Baltzopoulos V, Richards PJ, Maganaris CN (2011) Mechanical correction of dynamometer moment for the effects of segment motion during isometric knee-extension tests. *J Appl Physiol* 111:68–74
- Verschuren O, Ketelaar M, Takken T, Van Brussel M, Helders P, Gorter J (2008) Reliability of hand-held dynamometry and functional strength tests for the lower extremity in children with cerebral palsy. *Disabil Rehabil* 30:1358–1366
- Vimercati SL, Galli M, Rigoldi C, Ancillao A, Albertini G (2013) Feedback reliance during an arm-tapping task with obstacle avoidance in adults with Down syndrome. *Exp Brain Res* 226:631–638
- Vismara L, Cimolin V, Galli M, Grugni G, Ancillao A, Capodaglio P (2016) Osteopathic manipulative treatment improves gait pattern and posture in adult patients with Prader-Willi syndrome. *Int J Osteopath Med* 19:35–43
- Wagoner EV (1992) Apparatus for testing muscle strength. Patent: US5090421. Inventor: Wagoner, E. 5090421

- Wang C-Y, Olson SL, Protas EJ (2002) Test-retest strength reliability: hand-held dynamometry in community-dwelling elderly fallers. *Arch Phys Med Rehabil* 83:811–815
- Willemse L, Brehm MA, Scholtes VA, Jansen L, Woudenberg-Vos H, Dallmeijer AJ (2013) Reliability of isometric lower-extremity muscle strength measurements in children with cerebral palsy: implications for measurement design. *Phys Ther* 93:935–941
- Wuang Y, Chang J, Wang M, Lin H (2013) Test–retest reliabilities of hand-held dynamometer for lower-limb muscle strength in intellectual disabilities. *Res Dev Disabil* 34:2281–2290

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