Analysis of Knee Strength Measurements Performed by a Hand-Held Multicomponent Dynamometer and Optoelectronic System

Andrea Ancillao, Stefano Rossi, and Paolo Cappa

***Abstract*— The quantification of muscle weakness is useful to evaluate the health status and performance of patients and athletes. In this paper, we proposed a novel methodology to investigate and to quantify the effects induced by inaccuracy sources occurring when using a hand-held dynamometer (HHD) for knee strength measurements. The validation methodology is based on the comparison between the output of a one-component commercial HHD and the outputs of a six-component load cell, comparable in dimension and mass. An optoelectronic system was used to measure HHD positioning angles and displacements. The setup allowed to investigate the effects induced by: 1) the operator's ability to place and to hold still the HHD and 2) ignoring the transversal components of the force exchanged. The main finding was that the use of a single component HHD induced an overall inaccuracy of 5% in the strength measurements if the angular misplacements are kept within the values found in this paper ($\leq$15°) and with a knee range of motion $\leq$22°. Extension trials were the most critical due to the higher force exerted, i.e., 249.4 ± 27.3 versus 146.4 ± 23.9 N of knee flexion. The most relevant source of inaccuracy was identified in the angular displacement on the horizontal plane.**

***Index Terms*— Hand-held dynamometer (HHD), knee strength, load cell, lower limb, quality assurance, strength measurements.**

I. Introduction

Muscle strength measurement that is the evaluation of the maximum force produced by voluntary contraction of muscles is useful in clinical decision making and for the functional evaluation of patients and sportsmen. Strength measurements are also widely applied to test the effectiveness of training and rehabilitation programs [1]–[4]. For example, strength measurements were involved in the assessment of complex motor functions such as advanced features of gait [5] and treadmill training [6]. The measurement of the maximum exerted force allows the estimation of joint moment, which in turn helps the assessment of the healthiness of tendons and ligaments, and, consequently, the stability of the joint itself [7]. In a clinical perspective, some musculoskeletal or genetic diseases are characterized by gait and muscular disorders that are due to poor joint stability and muscle-tendon weakness [7]–[9]. Strength assessment is performed over time on the same subject in the pathologies affecting daily tasks, such as raising from a chair or walking [10]. Considering the inherent relation between muscle strength and muscle power, strength assessment plays an important role for the study of the previously cited pathologies and definition of clinical treatments [10].

The previously cited applications show the strong importance of measurement of force in clinical environments. Thus, a variety of quantitative methods, clinically feasible, were developed to assess human volitional muscle strength in cooperative patients. The earliest methods were based on the indirect measurement of muscle force and fatigue as the chair-stand test, but direct methods were generally to be preferred [2], [11] and they consisted in gathering the strength measurement via force sensors, often connected to *ad hoc* designed mechanical systems.

Nowadays, the gold-standard and widespread direct method is the isokinetic dynamometer, which is also commercially available [12]–[15]. It allows the gathering of force–velocity curves and the estimation of the maximum force exerted by the patient during a specified exercise. The isokinetic dynamometer showed high interrater and intrarater reliability and reproducibility for the measurement of joint forces and moments when it was applied to subjects of different ages, both on lower and upper limbs [3], [12], [16]. The isokinetic dynamometer was also used in combination with 3-D motion capture to obtain accurate joint moment measurements [17]. The main drawbacks are that the system is expensive and cumbersome, not portable, and it requires long time to prepare the subject. Thus, nowadays, a simpler and more clinically feasible method was developed. It is based on the use of the hand-held dynamometer (HHD) that is a small and portable single-component load cell that can be held in hand by an operator that applies it on defined anatomical landmarks [12], [16]. The HHD is a low-cost device, portable, easy to handle, and it does not require time-consuming procedures; for this reason, it is widely used in today's clinical practice.

Two HHD-based measurement methodologies were developed [18]: 1) the "make" test in which the examiner holds the dynamometer in a fixed position, while the subject exerts a maximal force against it and 2) the "break" test in which the

Manuscript received July 21, 2016; revised October 14, 2016; accepted October 18, 2016. Date of publication November 11, 2016; date of current version December 7, 2016. This work was supported by the MD-Paedigree European Project within FP7 (Co-PI: Paolo Cappa). The Associate Editor coordinating the review process was Dr. Amitava Chatterjee.

A. Ancillao is with the Department of Mechanical and Aerospace Engineering, Sapienza University of Rome, 00184 Rome, Italy.

S. Rossi is with the Department of Economics and Management–Industrial Engineering, University of Tuscia, 01100 Viterbo, Italy.

P. Cappa, deceased, was with the Department of Mechanical and Aerospace Engineering, Sapienza University of Rome, 00184 Rome, Italy.

Color versions of one or more of the figures in this paper are available online at http://ieeexplore.ieee.org.

Digital Object Identifier 10.1109/TIM.2016.2620799

examiner holds the HHD in place overpowering the maximal force exerted by the subject and then moving the participant's limb. The two methods were comparatively examined by Bohannon [18] concluding that both are reliable and repeatable if the examiner is able to promptly counteract the force exerted by the patient, while Phillips *et al*. [19] concluded that in the "break" test, some examiners could not complete the test due to their weakness, and then the "make" test had to be preferred. Due to its working principle, the use of HHD has some potential issues that may affect its accuracy and repeatability. Therefore, the study of metrological characteristics of HHDs was object of some studies in recent years [14], [20]. Wuang *et al*. [20] focused the study on the strength measurement of lower limb muscles in children with intellectual disabilities. They concluded that the use of the HHD could be considered practical and easy for clinicians, but it had some critical issues related both to the operator's training and to the positioning of the dynamometer on the subject's limb. The influence of the operator was tested by Kim *et al.* [12] by comparing three setups: 1) with the HHD fixed to the distal tibia by a Velcro strap; 2) with the HHD held by the operator; and 3) with an isokinetic dynamometer, assumed as a reference. They found that fixed and nonfixed methods showed good interrater reliability and the higher reliability was reached in the fixed methods. Martin *et al.* [14] confirmed that HHD offers a feasible, inexpensive, and portable test of quadriceps muscle strength for use in healthy older people even though it underestimates the absolute quadriceps strength compared with the isokinetic dynamometer, particularly in stronger participants.

From the literature survey, it emerges that the operator's skill to hold the HHD in the right position and orientation represents the main critical issue on the quality of the strength data. Moreover, a further unaddressed limitation of HHD-based measurement is related to the acquisition of a single force component that is the force value projected on the HHD sensitive axis, and consequently, the transversal components of force and the moments are ignored.

Most of the previous studies were focused on the comparative examination of the HHD versus the gold standard isokinetic dynamometer. To the best of the authors' knowledge, no previous work proposed a methodology to directly measure the sources of inaccuracies occurring when an HHD is used to measure knee strength. Our work is focused instead on the identification and quantification of the main causes of uncertainties related to manual strength measurements, which are normally neglected by the clinical operators. In particular, we focused on the HHD wrong positioning and limb/operator motion during the knee strength measurement. Hence, the aim of this paper was the quantitative evaluation of the sources of inaccuracies occurring when a commercial HHD is used. Our validation methodology estimated the effective position of the anatomical segments and the relevance of the transversal force components that are exerted by the subjects during strength measurements. The quantification of sources of inaccuracies was estimated by comparing the HHD outputs with an experimental setup with better metrological performances. The setup consisted of an optoelectronic system (OS) and a six-component load cell.

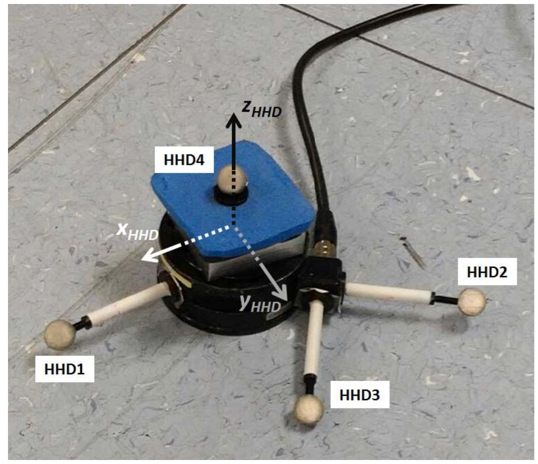

Fig. 1. Six-component load cell equipped with force transferring layers. Marker identification and the local reference system are also reported.

II. Materials and Methods

A. Subjects

Thirty healthy adult subjects were enrolled in the study: 18 males, 12 females, age 26.2 $\pm$ 2.1 years, height 173.6 $\pm$ 7.2 cm, and body mass 68.1 $\pm$ 8.7 kg. Subjects must not have had any neurological or orthopedic disorders and must not have had 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.

The choice of adult healthy subjects represents a worst case scenario, as high forces are exerted and the clinician must be strong enough to oppose the force of the subject.

B. Study Approval

This study complied with the principles of the Declaration of Helsinki, and it was approved by the Ethical Committee of the Children's Hospital "Bambino Gesù," Rome. Subjects were informed and signed consent before enrollment in the study.

C. Equipment

The force was measured by means of a commercial six-component HHD, i.e., the gamma F/T sensor (ATI Industrial Automation, USA). The load cell was equipped with an *ad hoc* aluminum force-transferring layer and a foam layer on the side that is in contact with the subject's limb (Fig. 1). The total thickness of the layers was 1.6 cm. Measurement full range was equal to 400 N for the force on the *z*-axis (principal axis), 130 N for the force on the transversal axes, and 10 Nm for the moment on each axis. Resolution was 1/20 N for the force and 1/800 Nm for the moment. The mass of the load cell was 0.255 kg, diameter 75.4, and height 33.3 mm. Load cell signal was amplified by means of the analog F/T controller, provided by the manufacturer, and streamed to the analog port of the Vicon system.

The lower limb motion was recorded by a Vicon MX OS (Oxford Metrics, U.K.), equipped with eight cameras. Data collection and the first processing, i.e., marker reconstruction and labeling, were conducted by the software Vicon Nexus 1.7 (200 Hz, Vicon Motion Systems, Oxford, U.K.). 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 the overall root-mean-square error (RMSE) of marker coordinates in 3-D space was less than 1 mm in a calibrated volume of about 2 m × 1 m × 2 m.

Force and moment data were synchronized with kinematic data recorded by the OS.

Ultimately, our setup allowed the identification of sources of inaccuracies in HHD measurements in order to give relevant information to the clinical personnel involved in the daily strength measurements improving their ability to perform such an evaluation. Specifically, the setup can quantify: 1) the actual dynamometer orientation with respect to the anatomical plane of the patient's lower limb; 2) the undesired motion of the limb that occurred during the strength measurements; 3) the actual forces and moments exchanged between the patient and the operator; and 4) the relevance of the transversal force components measured by the six-component load cell.

D. Motion Capture Protocol

The six-component HHD was equipped with four passive markers as shown in Fig. 1. 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 center with respect to the other markers that were used to build a local reference system (Fig. 1). The central marker was removed in the strength trials, and its position was reconstructed using a localization procedure based on the three fixed markers.

Subjects were equipped with an *ad hoc* defined marker protocol (Fig. 2) composed of 26 markers placed on the subject's skin surface. Landmarks were identified as follows: posterior and anterior iliac spines (four markers), lateral thighs (two markers), lateral and medial epicondyles (four markers), lateral and medial malleoli (four markers), lateral shanks (two markers), second metatarsal head (two markers), and calcaneous (two markers). Finally, two clusters of three markers (six markers) were applied on the thighs (in correspondence of the quadriceps femoris).

Positions of knee and ankle centers were computed as the midpoint between the two markers on epicondyles and on the malleoli, respectively. Hip center was reconstructed solving the plug-in-gait model, which is a modified version of the Davis protocol [21] when the subject is in the upright position. Instead, the position of hip center when the subject assumed the seated position was reconstructed by means of an optimal localization procedure since the markers on the pelvis were not always visible to the OS cameras.

A static trial was recorded for each subject before the trials for strength quantification. In the static trial, the subject was asked to stand up still in the OS calibrated volume for about 5 s. At the same time, HHD was kept still on the laboratory floor in the OS calibrated volume while zero input was applied. The static trial allowed evaluating: 1) the position and the orientation of local reference systems (LRSs) of each body segment and the HHD and 2) the offset signals gathered by the HHD.

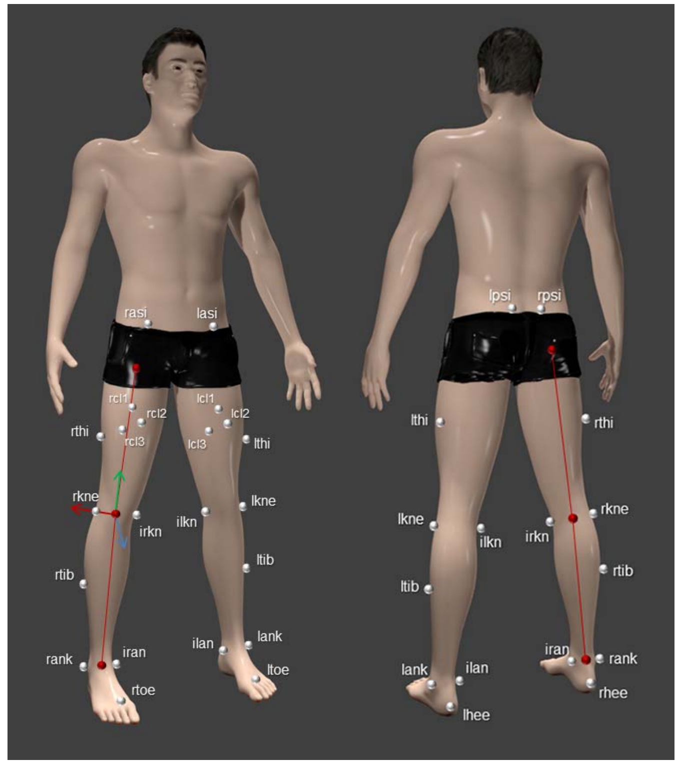

Fig. 2. Marker protocol used for the subjects: front and rear views. White dots are the markers used with the OS. Red dots are the reconstructed joint centers. Local reference system of the shank is shown in front view.

E. Strength Protocol

Testing was performed at the Movement Analysis and Robotic Laboratory of the Children's Hospital "Bambino Gesù."

Strength of the knee flexors and knee extensors muscles was measured by a protocol defined in a close cooperation with the clinical partners of the neuromuscular disease group within the MD-Paedigree project. This protocol consisted in a "make test" method [14], [18], [20]. The HHD was held by a trained clinician, and the trials were performed in accordance with [22]: subjects were sitting on a bench, and lower legs were hanging with hips and knees in flexion at about 90°. During the trials, in order to impede the movements of both the pelvis and the thigh, the subjects were stabilized by belts connected to the bench. The knee was left free to move, as flexion/extension was the measurement target. Hence, the HHD was placed proximally to the ankle on the anterior or posterior surface of the lower leg for the extension and flexion trials, respectively. The operator had to counteract subject's force by maintaining the limb and the HHD as still as possible.

The participants were instructed to exert the maximal force against the HHD for about 5 s while the operator counteracted the force trying to keep the shank still. The participants were also instructed to avoid explosive contraction, and they

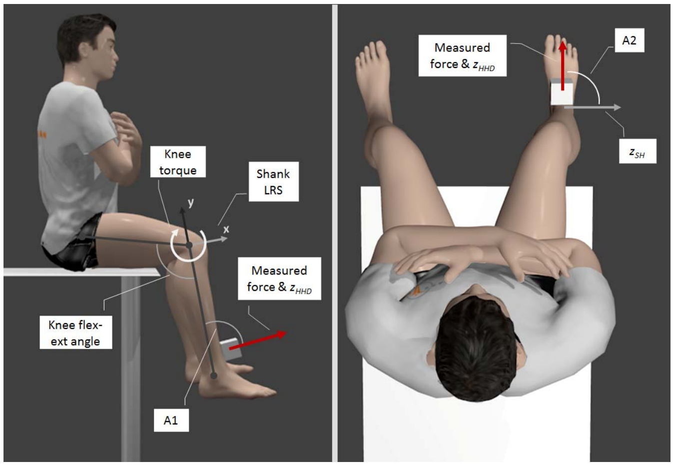

Fig. 3. Reference configuration with local reference system and computed parameters. Left: lateral view. Right: top view.

were instructed to increase force gradually from zero to the maximum achievable value [20].

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.

F. Data Analysis

Data processing algorithms were implemented in MATLAB (MathWorks, USA). Data processing consisted in: 1) the identification, via the markers, of LRSs for each body segment and HHD and 2) the evaluation of the selected kinematic and kinetic indices.

Two LRSs have been defined: for the shank, $\mathrm{LRS_{SH}}$ [Figs. 2 and 3], and for the HHD, $\mathrm{LRS_{HHD}}$ (Fig. 1). We defined for the $\mathrm{LRS_{SH}}$: 1) y_{SH}, the unit vector from ankle center to knee center, directed upward; 2) x_{SH}, the unit vector perpendicular to the plane defined by the knee medial and lateral epicondyles and the ankle center, and pointing forward; 3) z_{SH}, the unit vector perpendicular to x_{SH} and y_{SH}; and 4) the origin, located at the knee center. As concerns $\mathrm{LRS_{HHD}}$, we defined: 1) vmkr, the virtual marker defined as the projection of HHD4 on the plane represented by HHD1, HHD2, and HHD3; 2) x_{HHD}, the unit vector from vmkr to HHD1; 3) z_{HHD}, the unit vector perpendicular to the plane defined by HHD1, HHD2, and HHD3, pointing upward; 4) y_{HHD}, defined as cross product between z_{HHD} and x_{HHD}; and 5) origin, that is the virtual marker on the line between vmkr and HHD4 with an offset from HHD4 of 2.7 cm, which is the sum of thickness of the force coupling layers and the marker's radius.

G. Indices

In order to quantitatively study the quality of strength measurements, we defined some indices that are computed over the kinematic and kinetic data. The kinematic indices were defined as follows.

1) Range of motion (RoM) of knee angle is defined as the difference between the maximum and minimum of knee angle measured throughout the trial. Specifically, the knee angle was defined as the angle between the vector from knee center to hip center and the vector from knee center to ankle center [23]. RoM is an index to quantify the quality of strength measurements (low values of RoM represent higher adherence to the selected protocol, and actually the limb had to ideally remain still during the trial).
2) A_1 and A_2 represent the angles between z_{HHD} and y_{SH} and between z_{HHD} and z_{SH}, respectively (Fig. 3). A_1 and A_2 were evaluated at the instant in which the measurement of strength was gathered, that is, when the maximal force was recorded. A_1 and A_2 should be ideally equal to 90°, and their deviations from this value, δA_1 and δA_2, represent indices of incorrect positioning of HHD.

The previously defined indices were computed for both knee extension and knee flexion trials, and they were averaged between the five repetitions of each subject.

RoM and angular displacements are indices widely used in functional evaluation to quantify motion of limbs [23], [24]. In this paper, such indices were implemented to quantify the misplacements and displacements of HHD during measurements.

In order to assess repeatability of kinematic measurements, we computed the coefficient of variation (CV) for each parameter, addressed as $\mathrm{CV_{RoM}}$, $\mathrm{CV}_{\delta A1}$, and $\mathrm{CV}_{\delta A2}$. The CV was defined as the % ratio between the standard deviation (SD) and the mean value between the five repetitions for each subject.

As regards the kinetic data, we evaluated forces and moments acting on the knee joint center and reported in $\mathrm{LRS_{SH}}$ (${}^{\mathrm{SH}}\mathbf{F}$ and ${}^{\mathrm{SH}}\mathbf{M}$, respectively)

$$ {}^{\mathrm{SH}}\mathbf{F} = {}^{\mathrm{SH}}\mathbf{R}_{\mathrm{HHD}} \cdot {}^{\mathrm{HHD}}\mathbf{F} \tag{1} $$

$$ {}^{\mathrm{SH}}\mathbf{M} = {}^{\mathrm{SH}}\mathbf{R}_{\mathrm{HHD}} \cdot {}^{\mathrm{HHD}}\mathbf{M} + {}^{\mathrm{SH}}\mathbf{o}_{\mathrm{HHD}} \times {}^{\mathrm{HHD}}\mathbf{F} \tag{2} $$

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

We selected the following kinetic indices to quantify the information loss when the used HHD is a uniaxial load cell.

1) F_M is defined as the maximum value of ${}^{\mathrm{SH}}F_y$, which represents the strength measurement.
2) F_T is the transverse component of the force exerted by the subject. It allows the quantification of the intensity of lateral force that cannot be acquired with the usual clinical measurements conducted with a uniaxial load cell
$$ F_T = \sqrt{{}^{\mathrm{SH}}F_y^2 + {}^{\mathrm{SH}}F_z^2}. \tag{3} $$
3) M_M is defined as the maximum value of the knee flexion–extension moment, that is, ${}^{\mathrm{SH}}M_z$, when the strength measurement is conducted.
4) M_T is the transverse component of the knee moment. It represents the moment components not assessable if a uniaxial load cell was used to measure the knee moment
$$ M_T = \sqrt{{}^{\mathrm{SH}}M_x^2 + {}^{\mathrm{SH}}M_y^2}. \tag{4} $$

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, we also evaluated the nominal knee strength ($\hat{F}$) and the nominal knee moment ($\hat{M}$) as they are usually measured in clinical routine. Specifically, we considered that clinicians evaluate the force exerted by subjects with a uniaxial HHD and they estimate knee moment multiplying $\hat{F}$ by the distance between the HHD and the lateral malleolus 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 only ${}^{\mathrm{SH}}\mathbf{o}_{\mathrm{HHD}}$ that is the distance between the knee epicondyle and the HHD positioning as the nominal lever arm for the evaluation of knee moment

$$\hat{F} = {}^{\mathrm{HHD}}F_z \tag{5}$$

$$\hat{M} = {}^{\mathrm{SH}}\mathbf{o}_{\mathrm{HHD}} \cdot \hat{F}. \tag{6}$$

The inaccuracies related to both the strength and the knee moment were evaluated as the RMSE of the differences between actual values (F_M and M_M) and the nominal ones ($\hat{F}$ and $\hat{M}$); RMSEs were also normalized at the maximum values of $\hat{F}$ and $\hat{M}$

$$\mathrm{RMSE}_F = \sqrt{\frac{\sum_{i=1}^{N}\left(F_M^i - \hat{F}_i\right)^2}{N}} \cdot \frac{100}{\max_i(\hat{F}_i)}[\%]$$

$$\mathrm{RMSE}_M = \sqrt{\frac{\sum_{i=1}^{N}\left(M_M^i - \hat{M}_i\right)^2}{N}} \cdot \frac{100}{\max_i(\hat{M}_i)}[\%] \tag{7}$$

where N is the number of repetitions of the trials. Thus, RMSE values permitted the overall quantification of strength inaccuracy in knee moment measurements performed in the clinical routine.

To assess measurement repeatability, we also computed the CV for $\hat{F}$ and $\hat{M}$, addressed as $\mathrm{CV}_{\hat{F}}$ and $\mathrm{CV}_{\hat{M}}$.

Finally, to give an overall quantification of the quality of strength measurement, we proposed a novel synthetic quality index Q_{index}. It was defined to take into account both the angular misplacements of HHD (δA_1 and δA_2) expressed as a percentage of 90° and the transverse component of moment (M_T) expressed as percentage of the maximum value of the knee moment (M_M). We considered the transverse component of moment 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

$$Q_{\mathrm{index}} = 100\left(1 - \sqrt{\left(\frac{\delta A_1}{90}\right)^2 + \left(\frac{\delta A_2}{90}\right)^2 + \left(\frac{M_T}{M_M}\right)^2}\right). \tag{8}$$

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

TABLE I
KINEMATIC INDICES

	Extension	**Flexion**	**t-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
$CV_{\delta A1}$ [%]	5.0 (3.7)	2.1 (1.2)	0.002 *
$CV_{\delta A2}$ [%]	4.8 (2.4)	3.3 (1.3)	0.013 *

Mean (SD) values for the kinematic indices measured for the knee extension and flexion. * indicates a statistically significant difference between extension and flexion ($p<0.05$). The p values are reported in the t-test column.

H. Statistics

Descriptive statistic was 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 categorization for the Pearson coefficient r was considered, as suggested in [25]: $|r| = 1$: 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; and $|r| = 0$: zero.

III. Results

The results are reported in Tables I and II as means and SDs among the subjects included in the study; p-values of t-test are also reported. The Shapiro–Wilk test showed that kinetic and kinematic indices had a normal distribution, and therefore the selected statistical tests were applied.

The results of kinematic parameters are reported in Table I. The observed RoM, i.e., the angular variation of the knee flexion/extension angle across the trial, was never close to 0° and was statistically higher for the extension movement than the extension one. Moreover, the SD of RoM was higher for knee extension.

Positioning error δA_1 showed comparable values between knee flexion and extension. Instead, δA_2 assumed higher values for knee flexion. Values of δA_2 were always higher than δA_1 for both flexion and extension. As regards the coefficients of variation, no differences between flexion and extension were observed for the $\mathrm{CV}_{\mathrm{RoM}}$. It showed higher values than the coefficients of variation obtained for the positioning errors, i.e., $\mathrm{CV}_{\delta A1}$ and $\mathrm{CV}_{\delta A2}$. Both $\mathrm{CV}_{\delta A1}$ and $\mathrm{CV}_{\delta A2}$ were found statistically different between flexion and extension. The results of kinetic indices are reported in Table II.

TABLE II

KINETIC INDICES

	Extension	Flexion	t-test
$\hat{F}$ [N]	249.4 (27.3)	146.4 (23.9)	<< 0.001 *
$\hat{M}$ [Nm]	88.4 (12.4)	49.9 (8.3)	<< 0.001 *
F_M [N]	242.5 (28.8)	145.1 (26.4)	<< 0.001 *
F_T [N]	63.0 (22.8)	37.5 (14.1)	<< 0.001 *
M_M [Nm]	87.8 (12.5)	48.7 (9.5)	<< 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 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).

TABLE III

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 ($0.4 \leq |r| \leq 0.6$), ** a strong correlation ($0.7 \leq |r| \leq 0.9$).

Force and moment parameters $\hat{F}$, $\hat{M}$, F_M, F_T, M_M, and M_T were found significantly higher for knee extension than the knee flexion. All RMSE values were less than 5% (except for RMSE_M of knee flexion that was slightly higher).

A statistical significant difference was observed for RMSE_M of knee flexion that was higher than knee extension. RMSE_F was lower than RMSE_M, and no significant differences were observed between the two rotations. Considering the coefficients of variation, $\text{CV}_{\hat{F}}$ and $\text{CV}_{\hat{M}}$ were always lower than 10%. Both CVs were higher for knee extension; anyway, no significant differences were observed. Average Q_index was relatively high for both extension and flexion. It was slightly lower for knee flexion. The difference was close to significance.

The results of correlation analysis 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) are reported in Table III. 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, and RMSE_F parameters. Correlation was also strong between δA_1 and F_T and RMSE_F, while 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, while it was moderate with the other parameters.

IV. DISCUSSION

The main aim of this paper was to quantify the quality of the knee strength measurements by analyzing the inaccuracies relating to the operator and to the measurement system. More precisely, we analyzed the effects induced on the strength measurement by: 1) the therapist ability both in positioning the HHD and in keeping it still in place and 2) the current use of a single-component HHD that does not collect both moments and transversal components of force. Moreover, we studied the sources of inaccuracy occurring in the strength measurements by analysing their dependence with the kinematic parameters of HHD misplacement. Healthy adult subjects were enrolled in the study because they represent the most aversive case for the operator due to the relatively high force they can exert.

A. Dependency of the Strength Measurements on the Operator's Ability

In order to test the operator-dependent inaccuracies, when performing a strength measurement as a "make test" method, we measured the knee RoM that should be ideally close to 0°, as the subject's limb should remain still across the trial.

Analyzing the RoMs and the CV_RoM, values higher than 0° were measured for both extension and flexion. Moreover, knee RoM was statistically higher for extension trials, in which the force exerted was higher, than flexion ones (Table II), indicating that the operator was not able to completely counteract the participant's force, and therefore he was not able to hold the limb completely still with poor repeatability across the trials.

Considering the errors in the HHD placement, 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 trials. The index variability was relatively low, indicating a good repeatability in the angular positioning of 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 subject's ankle in knee flexion trials. It follows that the operator has to pay a special attention to avoid the rotation of the HHD on the horizontal plane.

Considering the coefficients of variation CV of the outputs of the strength measurements ($\text{CV}_{\hat{F}}$ and $\text{CV}_{\hat{M}}$), a low level of variability was observed ($\leq 5\%$) showing a good intra-subject repeatability of measurements for both knee flexion and knee extension trials, in agreement with the literature outcomes [12], [14], [19]. Moreover, $\text{CV}_{\hat{F}}$ and $\text{CV}_{\hat{M}}$ 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 this paper were higher during the extension trial than the flexion one implying a greater difficulty for the operator to maintain still the participant's limb. This finding is in agreement with the results in [26] that showed higher quality of the trials achieved by fixing the HHD in contrast to HHD freely held by the operator.

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.

B. Comparison Between Six-Component and One-Component HHD

The inaccuracy associated with 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 us to quantify the accuracy of strength and knee 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, 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 data set acquired in this paper.

In order to synthetically describe the quality of a strength measurement, according to the previously discussed parameters, we computed a synthetic quality index Q_{index}. The average of Q_{index} value was 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. The Q_{index} is also useful "on the field" applications, as it provides an overall quantification of the quality of a single trial and may help the clinician to identify trials to be discarded.

C. Correlation Between Operator's Ability and Strength Measurement Accuracy

Correlation between kinematic and kinetic parameters was analyzed 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 the misplacement indices δA_2 and δA_1 and F_T, M_T, and RMSE_F, while the correlation was low with RMSE_M. It can be stated that the angular misplacements had effect on the lateral undesired components of force and moment, while the error on the actual moment was not affected by an incorrect orientation of HHD. The misplacements influenced in a greater extent the force measurement than the moment one. 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 paper, does not affect the measurements in terms of lost information due to lateral components, which are not measured by commercial HHDs in clinical practice.

Concerning knee flexion trials that are characterized by a lower exerted force, correlations were observed only between δA_2 and all the parameters. Specifically, the correlation was strong only versus M_T, while it was moderate toward the other parameters. The strong correlation between δA_2 and M_T, also observed for the extension trial, demonstrated that the main misplacement of the HHD is on the horizontal plane and an increase in δA_2 could drastically 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 pay attention 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 does not affect the lateral components of force and moment if it is maintained within the values found in this paper.

In conclusion, the HHD misplacement quantified via the δA_2 index appears to be the main critical parameter for the quality assessment of a strength measurement.

V. Conclusion

This paper investigated the effects of sources of inaccuracies occurring when using a single-component HHD. The validation methodology is based on the concurrent use of a six-component HHD and an OS. This paper showed that the limb of subjects did not remain perfectly still during knee strength measurements and it represents a nonnegligible source of inaccuracy. From our measurements, we concluded that the use of uniaxial HHDs can be assumed as reliable and accurate enough for both knee flexion and extension, if lower limb displacements and HHD misplacement are kept within the values found in this paper (see Table I). The more critical measurements are in the knee extension, where the most affecting index was the HHD angular incorrect positioning on the horizontal plane.

According to the data collected for this paper, 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 results here discussed may lead to a better understanding of HHD measurements and provide directions to the 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.

ACKNOWLEDGMENT

The authors would like to thank the clinical staff of Children's Hospital "Bambino Gesù" of Rome, IT, for the use of their lab and for the support provided in data acquisition.

CONFLICT OF INTEREST

The authors declare that they have no conflict of interest.

REFERENCES

[1] R. W. Bohannon, "Hand-held compared with isokinetic dynamometry for measurement of static knee extension torque (parallel reliability of dynamometers)," *Clin. Phys. Physiol. Meas.*, vol. 11, no. 3, pp. 217–222, 1990.

[2] G. M. Allen, S. C. Gandevia, and D. K. McKenzie, "Reliability of measurements of muscle strength and voluntary activation using twitch interpolation," *Muscle Nerve*, vol. 18, no. 6, pp. 593–600, 1995.

[3] V. A. Hughes *et al.*, "Longitudinal muscle strength changes in older adults influence of muscle mass, physical activity, and healt," *J. Gerontol. A, Biol. Sci. Med. Sci.*, vol. 56, no. 5, pp. B209–B217, May 2001.

[4] R. J. Maughan, J. S. Watson, and J. Weir, "Strength and cross-sectional area of human skeletal muscle," *J. Physiol.*, vol. 338, pp. 37–49, May 1983.

[5] J. P. Ferreira, M. M. Crisostomo, and A. P. Coimbra, "Human gait acquisition and characterization," *IEEE Trans. Instrum. Meas.*, vol. 58, no. 9, pp. 2979–2988, Sep. 2009.

[6] K.-H. Chen, J. Fang, and S.-W. Yeh, "Design of an unobtrusive reaction force measurement and its application on treadmills," *IEEE Trans. Instrum. Meas.*, vol. 61, no. 7, pp. 1955–1965, Jul. 2012.

[7] R. Brunner and E. Rutz, "Biomechanics and muscle function during gait," *J. Children's Orthopaedics*, vol. 7, no. 5, pp. 367–371, Nov. 2013.

[8] M. Galli *et al.*, "The effects of muscle hypotonia and weakness on balance: A study on Prader–Willi and Ehlers–Danlos syndrome patients," *Res. Develop. Disab.*, vol. 32, no. 3, pp. 1117–1121, 2011.

[9] A. Ancillao, M. Galli, C. Celletti, M. Castori, G. Albertini, and F. Camerota, "Temporomandibular joint mobility in adult females with Ehlers-Danlos syndrome, hypermobility type (also known as joint hypermobility syndrome)," *J. Cranio-Maxillary Diseases*, vol. 1, no. 2, pp. 88–95, Jul. 2012, doi: 10.4103/2278-9588.105697.

[10] P. Capodaglio, L. Vismara, F. Menegoni, G. Baccalaro, M. Galli, and G. Grugni, "Strength characterization of knee flexor and extensor muscles in Prader–Willi and obese patients," *BMC Musculoskeletal Disorders*, vol. 10, no. 1, p. 47, 2009.

[11] C. J. Jones, E. R. Rikli, and W. C. Beam, "A 30-s chair-stand test as a measure of lower body strength in community-residing older adults," *Res. Quart. Exerc. Sport*, vol. 70, no. 2, pp. 113–119, 1999.

[12] W. K. Kim, D. K. Kim, K. M. Seo, and S. H. Kang, "Reliability and validity of isometric knee extensor strength test with hand-held dynamometer depending on its fixation: A pilot study," *Ann. Rehabil. Med.*, vol. 38, no. 1, pp. 84–93, Feb. 2014.

[13] D. E. Tsaopoulos, V. Baltzopoulos, P. J. Richards, and C. N. Maganaris, "Mechanical correction of dynamometer moment for the effects of segment motion during isometric knee-extension tests," *J. Appl. Physiol.*, vol. 111, no. 1, pp. 68–74, Jul. 2011.

[14] H. J. Martin, V. Yule, H. E. Syddall, E. M. Dennison, C. Cooper, and A. A. Sayer, "Is hand-held dynamometry useful for the measurement of quadriceps strength in older people? A comparison with the gold standard Biodex dynamometry," *Gerontology*, vol. 52, no. 3, pp. 154–159, Jan. 2006.

[15] J. C. Janssen and L. Le-Ngoc, "Intratester reliability and validity of concentric measurements using a new hand-held dynamometer," *Arch. Phys. Med. Rehabil.*, vol. 90, no. 9, pp. 1541–1547, Sep. 2009.

[16] M. L. Fulcher, C. M. Hanna, and C. R. Elley, "Reliability of handheld dynamometry in assessment of hip strength in adult male football players," *J. Sci. Med. Sport*, vol. 13, no. 1, pp. 4–80, Jan. 2010.

[17] A. Van Campen, F. De Groote, I. Jonkers, and J. De Schutter, "An extended dynamometer setup to improve the accuracy of knee joint moment assessment," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 5, pp. 1202–1208, May 2013.

[18] R. W. Bohannon, "Make tests and break tests of elbow flexor muscle strength," *Phys. Therapy*, vol. 68, no. 2, pp. 193–194, Feb. 1988.

[19] B. A. Phillips, S. K. Lo, and F. L. Mastaglia, "Muscle force measured using 'break' testing with a hand-held myometer in normal subjects aged 20 to 69 years," *Arch. Phys. Med. Rehabil.*, vol. 81, no. 5, pp. 653–661, May 2000.

[20] Y. P. Wuang, J. J. Chang, M. H. Wang, and H. C. Lin, "Test–retest reliabilities of hand-held dynamometer for lower-limb muscle strength in intellectual disabilities," *Res. Develop. Disab.*, vol. 34, no. 8, pp. 2281–2290, 2013.

[21] R. B. Davis, S. Õunpuu, D. Tyburski, and J. R. Gage, "A gait analysis data collection and reduction technique," *Human Movement Sci.*, vol. 10, no. 5, pp. 575–587, 1991.

[22] M. N. Eek, A. K. Kroksmark, and E. Beckung, "Isometric muscle torque in children 5 to 15 years of age: Normative data," *Arch. Phys. Med. Rehabil.*, vol. 87, no. 8, pp. 1091–1099, 2006.

[23] S. L. Vimercati, M. Galli, C. Rigoldi, A. Ancillao, and G. Albertini, "Feedback reliance during an arm-tapping task with obstacle avoidance in adults with down syndrome," *Experim. Brain Res.*, vol. 226, no. 4, pp. 631–638, May 2013.

[24] S. Vimercati, M. Galli, C. Rigoldi, A. Ancillao, and G. Albertini, "Motor strategies and motor programs during an arm tapping task in adults with down syndrome," *Experim. Brain Res.*, vol. 225, no. 3, pp. 333–338, 2013.

[25] C. Dancey and J. Reidy, *Statistics Without Maths for Psychology*. Harlow, U.K.: Pearson Education, 2004.

[26] B. A. Laing, F. L. Mastaglia, S. K. Lo, and P. Zilko, "Comparative assessment of knee strength using hand-held myometry and isometric dynamometry in patients with inflammatory myopathy," *Physiotherapy Theory Pract.*, vol. 11, no. 3, pp. 151–156, Jan. 1995.

Andrea Ancillao received the master's degree in clinical and biomedical engineering from the Sapienza University of Rome, Rome, Italy, in 2010, where he is currently pursuing the Ph.D. degree in industrial production engineering.

From 2011 to 2013, he was as a Research Assistant with the Politecnico di Milano University, Milan, Italy, and IRCSS San Raffaele Pisana, Rome. His current research interests include biomechanics, functional evaluation, and the design of measurement protocols for the analysis of human motion and rehabilitation.

Stefano Rossi received the Laurea degree (Hons.) in mechanical engineering from the Sapienza University of Rome, Rome, Italy, in 2004, and the Ph.D. degree in mechanical and thermal measurements with a specialization in measurements in biomechanics from the University of Padua, Padua, Italy, in 2008.

He is currently an Assistant Professor with the Department of Economics and Management, Industrial Engineering, University of Tuscia, Viterbo, Italy. His current research interests include biomechanics, design of robotic devices for rehabilitation robots, and the analysis of performances of measurement systems for the analysis of human movements.

Paolo Cappa received the Laurea degree (Hons.) in mechanical engineering from the Sapienza University of Rome, Rome, Italy, in 1980.

He was a Full Professor with the Department of Mechanical and Aerospace Engineering, Sapienza University of Rome. He was also an Adjunct Professor with the Department of Mechanical and Aerospace Engineering, New York University Tandon School of Engineering, New York, NY, USA.