# Measurement device for ankle joint kinematic and dynamic characterisation

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**Abstract**—*The paper describes a measurement device for obtaining the kinematic characterisation and isometric loading of ankle joints under different working conditions. Non-invasive,* in vivo *experiments can be conducted with this experimental apparatus, the potential of which could be usefully exploited in basic biomedical research, prosthesis design, clinical applications, sports medicine and rehabilitation. The device determines the 3D movement of the foot with respect to the shank and evaluates the torques and moments around the three articular axes in relation to any desired angular position of the ankle complex. When integrated with superficial electromyographic techniques and electrical stimulation, it allows the assessment of the functionality of the lower leg in both mechanical and myo-electrical terms. The paper reports the main mechanical and electronic features of the device (high linearity; maximum moment ranges ±300 Nm for flexion–extension, ±35 Nm for both pronation–supination and internal–external rotation; angular ranges: ±100° of dorsi-plantar flexion, ±50° of internal–external rotation and prono-supination; linear ranges: ±25 mm along each axis). Results from a healthy volunteer, under voluntary or stimulated conditions, helped in testing its operatability, reliability, robustness, repeatability and effectiveness. Preliminary simplified protocols have been also applied to 20 healthy volunteers, and the main results were 80.8 ± 11.9° of internal–external rotation, 46.2 ± 9.1° of prono-supination and 74.6 ± 13.1° of flexion–extension. Torques and moments were normalised with respect to a body mass index of 30. The maximum plantar flexion moment (57.5 ± 21.3 Nm) was measured with the foot at 15° of dorsal flexion; the maximum dorsal flexion moment (50.2 ± 20.3 Nm) was measured with the foot at 15° of plantar flexion.*

**Keywords**—*Foot–ankle kinematics, Isometric contraction, Ankle measurement device, In vivo muscle assessment*

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## 1 Introduction

VARIOUS BIOMECHANICAL models have been conceived quantitatively to analyse foot motion with respect to the shank and the related dynamics, in terms of forces, torques and moments exerted on the foot by the muscular groups of the lower leg (PROCTER and PAUL, 1982; WYNARSKY and GREENWALD, 1983).

These works have been based on *in vitro* and *in vivo* experiments. Most of the *in vivo* research methods are invasive and only give a partial and simplified description of the articular complex, which is often modelled as a two-hinge joints chain (DUL and JOHNSON, 1985; INMAN, 1976; SCOTT and WINTER, 1993).

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Even though many important issues still need to be discussed, several features have been generally observed and discussed elsewhere (LEARDINI, 2000). One of the most important findings was the changing of the instantaneous axis of rotation of the ankle (tibio-talar) joint, thus suggesting that the hinge joint complex is an oversimplification for the ankle joint. A close interaction was also claimed between the geometry of the ligaments and the shape of the articular surfaces in guiding and stabilising motion at the ankle joint.

As for the motion of the subtalar joint, even greater variability was found among the published experiments, mainly depending on conditions and methods of examination. The orientation of its axis of rotation was frequently discussed. As reported by LEARDINI (2000), several authors described the subtalar joint as a hinge joint, with an oblique axis running approximately in the antero-medio-superior to postero-latero-inferior direction (MANTER, 1941; INMAN, 1976; ISMAN and INMAN, 1969). Thus the motion of the calcaneus with respect to the talus involves rotations in all three anatomical planes. Further studies demonstrated the changing positions and orientations for this axis, and thus more complex models have been proposed, that consider the multi-axial nature of the joints (SANGEORZAN and SIDLES, 1995).

When the whole ankle complex was considered, the 'closed spatial kinematic chain' principle was described by HUSON (1984). On the basis of this principle, any movement of any tarsal bone occurs according to a fixed and coupled pattern that allows only one degree of freedom (DOF), and polyaxial joints should replace single axis of rotation joints. Ligaments play an important role in the guidance of motion, as the articular contact showed incongruity during movement (HUSON, 1984).

The complex mechanical coupling between the ankle and subtalar joints has also been compared with that of a universal-joint type of linkage, which is able to transfer prono-supination of the calcaneus to tibial internal–external rotation and *vice versa* (OLERUD and ROSENDHAL, 1987). The ankle complex has also been recently modelled as a system with two monocentric, single DOF hinge joints (SCOTT and WINTER, 1993). Alternatively, the kinematics of these joints has been modelled as an equivalent screw axis that provides a full six DOF (ENGSBERG and ANDREWS, 1987). Such representations provide information about the entire motion, but little information on normal or pathological motion of either joint.

Studies on the kinematics of the ankle joint have shown that, during the loaded movement of the foot with respect to the shank, it is difficult to predict the relative movements of the parts of the ankle complex (LEARDINI, 2001). Approximated models can be built by assuming that the articular axes are coincident at a fixed point, located in the middle of the segment, that connects the tips of the tibial and fibular malleoli (LUNDBERG, 1989; LUNDBERG *et al.*, 1989*a*; SIEGLER *et al.*, 1988). However, the development of these models is very difficult, and simpler models based on fixed hinges or spherical joints do not yield reality-fitting representations of the ankle joint complex (SIEGLER *et al.*, 1988; ENGSBERG, 1987).

Moreover, it has been observed that the ankle and the subtalar joint are not subject to independent regulation, and thus the resultant combined movement might be of greater functional value than the separate components (ELFTMAN, 1960).

Further difficulties arise in *in vivo* static and dynamic analyses, owing to the indeterminacy of the resultant force exerted on the foot. On the other hand, knowledge of the force and torque or moment components along and around the articular axes is important for estimation of the inner stresses of the single parts of the complex. This bulk of information could be very useful for various applications, such as the design of prostheses or the definition of rehabilitating therapies.

In this investigation, we developed a tool for the complete description of the kinematics and the isometric loading of the ankle joints (talo-crural and subtalar joints) under different conditions. Non-invasive, *in vivo* experiments can be conducted with this experimental apparatus. The device allows 6 DOF, three rotations and three translations; six transducers allow the movement of the foot with respect to the shank to be fully described. Then, for every position of the ankle articular complex, it is possible to evaluate the moments related to the movements of flexion–extension (in the sagittal plane), the torques related to internal–external rotation (in the transverse plane) and the moments related to prono-supination (in the frontal plane).

The characterisation of the articular complex can be completed by integration of the proposed device with advanced superficial electromyographic techniques and electrical stimulation. In this way, the functionality of the main muscular compartments of the lower leg can be assessed in both mechanical and myo-electrical terms. It is worth highlighting the usefulness of the ankle measurement system in clinical applications, to analyse the musculo-skeletal functionality of the ankle complex before and during the rehabilitation process. Besides this, when used under strictly controlled conditions, it may contribute to basic biomedical research.

This paper describes the main mechanical and electronic features of the device. For a better presentation of its potentialities, we give a preliminary, integrated experimental configuration and some relevant results from a healthy volunteer. The reported experiment helped to test the main functional characteristics of the whole measurement system in terms of operation, reliability, precision, accuracy, robustness and repeatability.

In a more extensive use of the device, it was applied to a group of 20 healthy volunteers, whose articular mobility and muscular functionality were investigated in terms of mean values and intra-class variability. Simplified protocols were applied to the subjects, without any electrical stimulation, and EMG electrodes were only used to monitor the contraction level before each voluntary contraction. The main results are reported and discussed in the study.

## 2 Materials and methods

### 2.1 *Mechanical description of the ankle measurement device*

The mechanical system was designed to detect foot displacements with respect to the shank and torques or moments at the ankle articular complex. It basically consists of an open loop, seven links and a six-DOF chain. The device and a detailed diagram of it are reported in Fig. 1. The seven links are

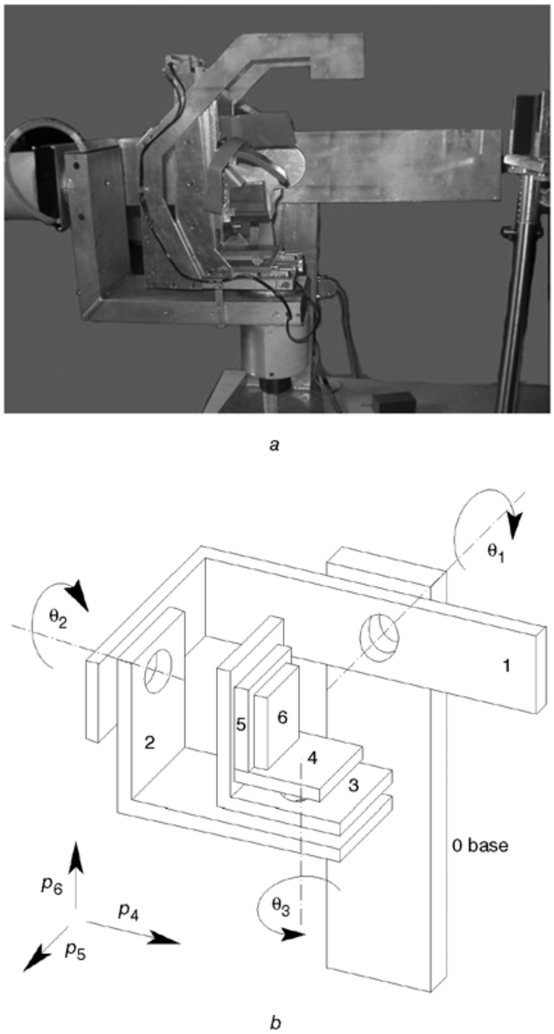



**Fig. 1** *(a) Ankle measurement device. (b) Diagram of the seven link, six-DOF mechanical chain. Link 0 is solid with the patient's shank; link 6 is solid with foot*

connected through three revolving joints for angular movements and three prismatic joints for linear movements. Measurements of torques are possible in any desired fixed position, by blocking of the revolving joints by means of an electromagnetic system. Angular movements are measured by three angular potentiometers solid with the revolving joints; linear movements are measured by three linear potentiometers solid with the prismatic joints.

The subject is seated on a three-DOF adjustable seat, and his/her foot is positioned and fixed to plate 6. The fore-foot is blocked by velcro stripes. Undesirable, relative movements between the rear-foot and plate are prevented by means of an aluminium block for the calcaneus, with the interposition of a polymethylsiloxane mould, prepared on the subject's calcaneus so as to reproduce accurately even small details of its surface. Extended or flexed, the leg is tied to a fork-shaped support rigidly fixed to the base link 0.

The mechanical layout has been designed so as to render the axes of the device as close as possible to the middle point of the segment connecting the tips of the medial and lateral malleoli. The axes of the anatomical reference system, solid with the shank and with the base link 0, result from the mutual intersections of the sagittal, transverse and frontal planes. The three axes of the foot reference system, which is solid with the device's final link (link 6), coincide with the anterior–posterior axis of the foot (axis of pronation–supination), with the line connecting the tips of the medial and lateral malleoli (axis of dorsal–plantar flexion) and with the line perpendicular to plate 6 (axis of internal–external rotation) (LUNDBERG *et al.*, 1989*b*). With the transformation matrix $T$, computed over Denavit–Hartenberg notation (ASADA and SLOTINE, 1986), the position and orientation of the foot (reference system solid with the end link) are then related to the shank (reference system solid with the base link).

A detailed description of the applied methodology and related computations is reported in (BELFIORE *et al.*, 2001). Briefly, the general transformation matrix $T$ is obtained by multiplying standard transformation matrices ($A_i^{i-1}$) that link the generic reference system $i$ with the reference system $i-1$. End frame 6 is then related to first frame 0 through the following matrix:

$$T = A_1^0(\theta_1) \circ A_2^1(\theta_2) \circ A_3^2(\theta_3) \circ A_4^3(p_4) \circ A_5^4(p_5) \circ A_6^5(p_6)$$

where $\theta_1, \theta_2, \theta_3$ stand for rotations of the leg around the flexion–extension, internal–external and pronation–supination articular axes, respectively, and $p_4$, $p_5$, $p_6$ represent compression–distraction, medio-lateral and anterior–posterior translations. Two assumptions have been made

(i) small constructive errors in the alignment of the device links were considered negligible
(ii) in the initial configuration, fixed and mobile reference systems were assumed to coincide.

Denavit–Hartenberg parameters were accurately determined using optical methods, with a direct reading precision of 0.01°*.

A further transformation is then mandatory to determine the kinematics of the foot frame in the Cartesian space, as the matrix $T$ relates the vector of joint variables to the displacements and rotations of the foot. The Jacobian matrix of the mechanism (matrix $J$), which maps the six-dimensional joint rate vector onto the six-dimensional vector of Cartesian velocities, is then calculated. Thanks to the duality of kinematics and statics, the transpose of the same matrix $J$ relates joint torques to generalised force exerted over link 6.

*WILD Theodolite T2mod, WILD HEERBRUGG, Switzerland

### 2.2 *Device electronic features*

Torques are measured through three extensometric bridges† mounted so as to transduce the torsional strain of the shafts of the three revolving joints. Bridge excitation was established at 5 V as a trade-off between power consumption and sensitivity. Sensitivity has been further enhanced by use of full-bridge configuration, which also guarantees minimum offset thermal drift. The transducers have been positioned as far as possible from the discontinuities of the shaft section, to produce a linear stress–strain relationship. Connecting cables were properly soft-soldered, so as not to transmit potential cable tension to the strain gauge. Amplifier gains were adjusted to obtain ranges in agreement with the literature (BOBBERT and VAN INGEN SCHENAU, 1990): ±250 Nm for flexion–extension moment, and ±20 Nm for both pronation–supination moment and internal–external torque. To improve resolution at lower values, two further gains can be set for each amplifier, when needed.

The joint rotation angles are measured by means of continuous rotation precision potentiometers‡ input voltage 10 V, sensitivity 0.028 V degree$^{-1}$, linearity tolerance ±2%), with elements in conductive plastic and bushing mount format. Translations are measured with linear potentiometers** (input voltage 10 V).

The potentiometer outputs are connected to voltage-followers to produce low-impedance voltage signals. Offsets were regulated to set zero-voltage output in relation to the initial reference position of the ankle joints (90° between foot and shank in the sagittal plane, 0° in the other two planes; knee 90° flexed to have the shank rotated by 20° backwards with respect to the perpendicular to the ground). Amplifiers gains were properly set to measure, in relation to maximum output range (±5 V): ±100° in dorsi–plantar flexion, ±50° in internal–external rotation and ±50° in prono–supination.

Moment or torque transducers were accurately calibrated by the application of known moments. For each transducer calibration, a proper configuration of the mechanical chain was used theoretically to nullify the moments due to gravity, and therefore calibrating moments were composed of a known force component (measured by IMADA dynamometer, range 0–1000 N, resolution 1 N) applied at a known lever arm. Sensitivity was then assessed for each torque at each possible amplifier gain.

The nine voltage signals were low-pass filtered at 10 Hz before being sampled and acquired by means of the A/D converter†† (resolution 12 bits, input range 10 V). This specific board was chosen because of its capacity to acquire up to 64 single-ended analogue signals. Thus the ankle measurement device can be used in conjunction with other measurement devices, such as multi-channel EMG amplifiers. The sampling rate ranged from 100 Hz (ankle device alone) to 1024 Hz (ankle device + EMG).

### 2.3 *Preliminary experimental configuration*

2.3.1 *Measurements on one healthy volunteer:* The measurements described in this Section have been taken on a healthy volunteer (male, 28 years, 75 kg, 185 cm) to assess the experimental set-up, test the feasibility of the associated measurement protocol and detect a reliable criterion for the normalisation of the torque and moment values acquired under electrically stimulated conditions.

The ankle measurement device was synchronised with a commercial EMG amplifier and a commercial electrical stimulator.

†Measurememt Group, Inc., Raleigh, NC, USA
‡Series 357-0-0-502, Spectrol
**Leane International, Parma
††ATMIO-64F5, National Instruments

The surface EMG device[††] uses particular electrode arrays where myo-electric signals are simultaneously captured by 16 unipolar or 15 differential, equally spaced electrodes. Besides computing the usual temporal and spectral parameters of an EMG signal, this device yields a good estimation of the velocity of conduction. Inter-electrode distances and array configurations can be chosen according to specific measurement needs. In the experimental set-up described, seven differential EMG signals were simultaneously captured along each of two investigated muscles; thus two electrode arrays were used, with eight contacts each and with 10 mm inter-electrode distance. The rationale behind this choice was the simultaneous investigation into the effects of stimulation on the directly stimulated muscle and the eventual cross-talk from an adjacent one.

The electrostimulator used in this study is a dedicated current generator* that provides pulse trains of programmable amplitude, shape, duration and frequency. Its peculiarity is the possibility to deliver stimulating currents through eight electrodes simultaneously; the stimuli amplitude and shape are selectable for each channel independent from the others. In the test described, only one shape was used for the impulses (bi-phasic shape), to provide well-established, controlled experimental conditions.

The volunteer's right shank was aligned to the first link of the mechanical chain, previously rotated by $+20^\circ$ with respect to the horizontal plane. In this way, the knee was flexed by $90^\circ$, and the subject could sit in a more comfortable position. His foot was blocked at $90^\circ$ with respect to the shank in the sagittal plane, and at $0^\circ$ in the other two planes (frontal and transverse). This specific configuration, which we shall refer to as the neutral position, guarantees the measured torques and moments to be directly referred to the anatomical reference system. A sequence, performed three times, consisted of maximum voluntary isometric contractions of 5 s each, followed by 90 s of rest. More specifically, the volunteer was asked to contract the following muscles in the reported order: tibialis anterior, peroneus longus, gastrocnemii and muscles acting in dorsal flexion, internal rotation, external rotation, pronation, supination. The subject was trained to activate separately

(*a*) tibialis anterior by simultaneously dorsal flexing, internal rotating and supinating the foot
(*b*) peroneus longus by plantar flexing, external rotating and pronating the foot (KAPANDJI, 1983).

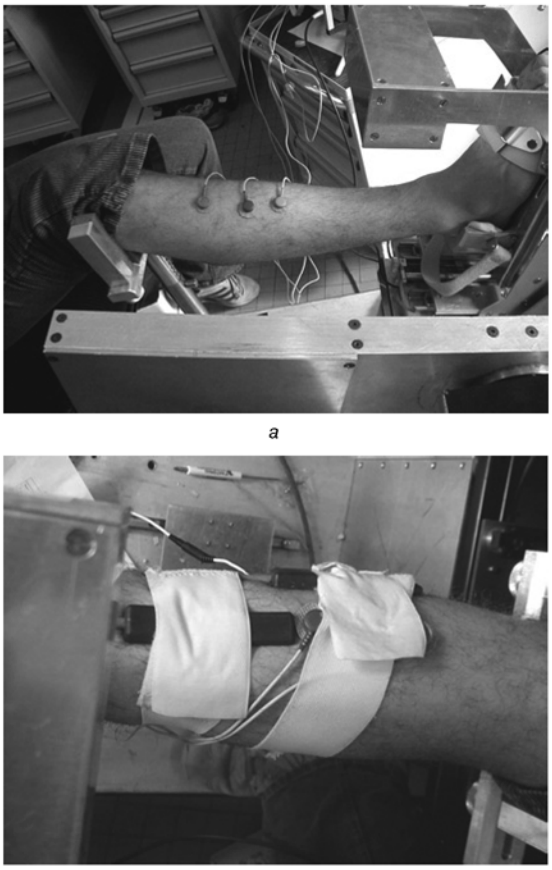
As regards angular excursions, the foot was first rigidly blocked in neutral position and then successively unlocked only along the direction of movement under measurement: maximum excursions in the directions of flexion–extension, internal–external rotation and prono-supination were acquired while the volunteer was performing slow cycles for a total duration of 5 s. Each acquired parameter was then averaged over the three repetitions.

To perform the stimulated contractions, the volunteer was blocked again in the neutral position, and three main motor points were identified on his tibialis anterior (Fig. 2). The most suitable stimulus amplitude was established for each motor point. Contractions were then induced by alternate stimulation of the three motor points in all possible combinations, one sequence at 25 Hz and another sequence at 90 Hz. A rest period of 90 s was observed between two successive contractions.

2.3.2 *Measurements on 20 healthy volunteers. Reference data:* Twenty healthy volunteers (age $50.4 \pm 14.6$ years, height $168.6 \pm 8.8$ cm, body mass $71.0 \pm 11.3$ kg) were recruited for a first investigation into ankle muscular functionality under well-controlled isometric conditions and ankle complex joint mobility. The recollection of the test subjects and objective clinical examination excluded serious neuro-musculo-skeletal pathologies in all of them that could influence their ankle functionality.

In the specific experimental set-up, EMG signals were simultaneously captured along each of two investigated muscles, namely the tibialis anterior and the lateral gastrocnemius; two electrode arrays were used, with eight contacts each and 10 mm inter-electrode distance. The rationale behind this choice was the monitoring of muscle activity before each contraction. The acquired data would also supply useful information about timing and level of muscle activation under different isometric configurations; however, the study of myo-electrical activity was beyond the scope of the present study, and the interpretation of the related data is not discussed in the following.

Each subject was positioned in neutral position, as described in section 2.3. He was first trained to perform the correct isometric contraction starting from rest. In this phase, the visual feedback of the EMG signals proved to be very helpful. The subject was then asked to perform a sequence of maximum isometric contractions, in both directions of each plane, to produce torques around the vertical axis, flexion–extension moments around the medio-lateral axis and pronation–supination moments around the longitudinal foot axis. Each contraction lasted 10 s and was followed by 90 s of rest. Flexion–extension moments were further investigated by the same exercise being repeated with the foot at $15^\circ$ and $30^\circ$ of plantar flexion and $15^\circ$ of dorsal flexion, respectively. Maximum torque and moment values were corrected with respect to the initial offset and then normalised with respect to body mass



a

b

**Fig. 2** *(a) Volunteer's positioning on measurement device. Electrical stimulation relating to three motor points of tibialis anterior. (b) Volunteer's complete equipment. Two electrode arrays have been added to acquire EMG signals from tibialis anterior and lateral gastrocnemius*

[††]ASE16, LISIN, Turin, Italy
*LISIN, Turin, Italy

index (BMI). This normalisation was chosen in the attempt to take into account, not only the dependence of muscle length and moment arm on the height of the subject, but also their eventual dependence on the mass of the bony structure. An ideal BMI of 30 was then introduced to express torques and moments as 'normalised' newton metres.

With regard to angular excursions, maximum excursions in the directions of flexion–extension, internal–external rotation and pronation–supination were acquired while the subject was performing slow cycles for a total duration of 10 s. The absolute maximum angular value in each direction was then identified and included in the study.

## 3 Results

### 3.1 *Characterisation of the ankle measurement device*

The main characteristics of the kinetic measurements are reported in Table 1. High linearity and overall accuracy were found in the desired torque ranges.

With regard to the kinematic measurements (rotations), the inaccuracies found were mainly due to non-linearities of the potentiometers that were close to those claimed by their manufacturers (±2% FS).

### 3.2 *Preliminary results*

3.2.1 *Measurements on one healthy volunteer:* Table 2 reports peak values of the torques and moments expressed during voluntary isometric contractions and of the angular excursions. Kinetic values were averaged over three maximum contractions; angular values were averaged over all cycles included in a 5 s performance. An acronym was added to each quantity to specify the axis around which it was applied (torque or moment expressed when the device is blocked) or performed (movement done when the device is allowed to rotate). More specifically, the indices were: f-e to refer to the medio-lateral axis (flexion–extension); i-e to refer to the longitudinal tibia axis (internal–external rotation); and p-s to refer to the longitudinal foot axis (prono-supination). Owing to the choice of the device reference system, the following measures are >0 for a right ankle complex: plantar-flexion moments, external rotation torques, supination moments and angular excursions in the directions of dorsal flexion, external rotation and pronation.

The values reported in Table 2 demonstrate that tibialis anterior and peroneus longus, when singularly activated, deliver similar dorsal-flexion moments ($\text{moment}_{\text{f-e}}$) and minor opposite values around the other two axes. When the two muscles are contracted simultaneously, as in the maximum voluntary dorsal flexion, the $\text{moment}_{\text{f-e}}$ is almost doubled, and the values around the other two axes almost compensate each other.

Table 3 reports peak values of torques and moments expressed under stimulated conditions. Electrical stimulation was alternatively applied in different combinations to the three motor points on the tibialis anterior, at 25 Hz and at 90 Hz, with bi-phasic shape and with amplitudes ranging from 45 to 48% of the maximum current (100 mA) available from each of the eight channels of the stimulator. The current amplitude was established for each motor point as the minimum amplitude required to detect a clear motor unit action potential (MUAP), without any interference.

As reported in Table 3, the simultaneous electrical stimulation of more than one motor point of the same muscle results in a moment that is almost the sum of the contributions of the

*Table 1 Main characteristics of torque and moment measurements (range, determination coefficient $R^2$ and maximum error) in 3 anatomical planes of reference system, for each of 3 possible gains*

| | Flexion–extension moment | | | Internal–external rotation torque | | | Prono-supination moment | | |
|---|---|---|---|---|---|---|---|---|---|
| | range, Nm | $R^2$ | maximum error, %FS | range, Nm | $R^2$ | maximum error, %FS | range, Nm | $R^2$ | maximum error, %FS |
| Minimum gain | ±300 | 0.997 | 0.6 | ±35 | 0.999 | 0.7 | ±35 | 0.997 | 0.7 |
| Medium gain | ±100 | 0.999 | 0.8 | ±15 | 0.995 | 0.9 | ±15 | 0.998 | 1.0 |
| Maximum gain | ±50 | 0.996 | 2.0 | ±7 | 0.997 | 2.1 | ±7 | 0.995 | 2.0 |

*Table 2 Torque and moment values and angular excursions averaged over three repetitions performed by healthy volunteer. Kinetic measurements were acquired during maximum voluntary isometric contractions of 5 s each. Angular excursions were acquired during slow cycles of 5 s each*

| | $\text{Moment}_{\text{f-e}}$, Nm | $\text{Torque}_{\text{i-e}}$, Nm | $\text{Moment}_{\text{p-s}}$, Nm |
|---|---|---|---|
| *Resisted actions* | | | |
| tibialis anterior activation | −20.2 ± 6.1 | −8.7 ± 1.9 | 5.8 ± 0.9 |
| peroneus longus activation | −20.8 ± 6.3 | 10.0 ± 2.5 | −4.0 ± 1.4 |
| plantar flexion (gastrocnemii) | 38.8 ± 5.5 | 2.0 ± 0.8 | 2.9 ± 0.9 |
| dorsal flexion | −34.2 ± 6.7 | 0.1 ± 0.1 | 1.3 ± 0.4 |
| internal rotation | −10.8 ± 3.3 | −12.8 ± 3.8 | 5.1 ± 1.4 |
| external rotation | −8.5 ± 2.1 | 11.6 ± 3.0 | 1.3 ± 0.4 |
| pronation | 4.9 ± 1.1 | 1.3 ± 0.4 | −5.6 ± 1.0 |
| supination | 4.2 ± 1.3 | 11.5 ± 2.2 | 10.0 ± 1.9 |
| | minimum, ° | maximum, ° | range, ° |
| *Angular excursions* | | | |
| flexion–extension | −50.0 ± 0.3 | 21.9 ± 1.0 | 71.9 ± 1.0 |
| internal–external rotation | −24.6 ± 1.8 | 44.7 ± 1.5 | 69.4 ± 3.1 |
| prono-supination | −23.4 ± 1.8 | 18.9 ± 0.6 | 42.3 ± 2.4 |

*Table 3 Torque and moment values obtained during stimulated isometric contractions of tibialis anterior. Each value has been averaged over three repetitions. Reference values obtained under maximum voluntary contractions (MVCs) are reported for comparison*

| Motor point | Frequency, Hz | $Moment_{f\text{-}e}$, Nm | $Torque_{i\text{-}e}$, Nm | $Moment_{p\text{-}s}$, Nm |
|---|---|---|---|---|
| MVC | | $-20.2 \pm 6.1$ | $-8.7 \pm 1.9$ | $5.8 \pm 0.9$ |
| 1 | 25 | $-8.3 \pm 0.8$ | $-0.7 \pm 0.03$ | $0.7 \pm 0.02$ |
| 2 | 25 | $-4.5 \pm 0.2$ | $-0.4 \pm 0.02$ | $0.4 \pm 0.01$ |
| 3 | 25 | $-3.7 \pm 0.2$ | $-0.4 \pm 0.02$ | $0.3 \pm 0.01$ |
| 1-2 | 25 | $-13.0 \pm 1.0$ | $-1.0 \pm 0.04$ | $1.0 \pm 0.04$ |
| 1-3 | 25 | $-9.1 \pm 0.8$ | $-0.3 \pm 0.03$ | $0.4 \pm 0.02$ |
| 2-3 | 25 | $-6.7 \pm 0.5$ | $-0.6 \pm 0.03$ | $0.7 \pm 0.03$ |
| 1-2-3 | 25 | $-13.2 \pm 1.1$ | $-0.7 \pm 0.01$ | $0.7 \pm 0.03$ |
| 1-2-3 | 90 | $-13.7 \pm 1.0$ | $-1.1 \pm 0.04$ | $1.1 \pm 0.03$ |

single motor points. The Table also shows that the peak values reached under 25 Hz stimulation are comparable with those under 90 Hz.

3.2.2 *Measurements on 20 healthy volunteers. Reference data:* Table 4 reports mean values and standard deviations of the maximum angular excursions measured in the three reference planes, in both directions.

Table 5 shows torque and moment values obtained during voluntary isometric contractions in both directions of the three planes. Pronation–supination moments and internal–external torques were only measured in neutral position. Flexion–extension moments were also measured with the foot at 15° and 30° of plantar flexion and 15° of dorsal flexion. Each value has been normalised with respect to a reference BMI of 30.

## 4 Discussion and conclusions

The paper describes the main mechanical and electronic features of a measurement device for the *in vivo* static characterisation of the foot–ankle complex.

From a mechanical point of view, the device has been completely characterised: reference systems based on the directions of the corresponding anatomical axes were associated with the link solid with the shank (first frame) and with the end link of the chain, which is solid with the foot. Adequate theory was

*Table 4 Mean values and standard deviations of angular excursions of ankle articular complex in three reference planes*

| Plane | Movement | Mean value, ° | Standard deviation, ° |
|---|---|---|---|
| transverse | internal rotation | 39.0 | 7.1 |
| | external rotation | 41.8 | 6.9 |
| | range | 80.8 | 11.9 |
| frontal | supination | 30.9 | 6.7 |
| | pronation | 15.3 | 5.9 |
| | range | 46.2 | 9.1 |
| sagittal | plantar flexion | 40.2 | 7.6 |
| | dorsal flexion | 34.4 | 8.0 |
| | range | 74.6 | 13.1 |

*Table 5 Torque and moment values obtained during voluntary isometric contractions in both directions of three planes. Pronation–supination moments and internal–external torques were only measured in neutral position. Flexion–extension moments were also measured with foot at 15° and 30° of plantar flexion, and 15° of dorsal flexion. Values listed are measured value × subject's BMI/30*

| Plane | Moment | Position | Mean value, Nm | Standard deviation, Nm |
|---|---|---|---|---|
| transverse | internal | neutral | 16.7 | 8.2 |
| | external | neutral | 19.6 | 7.8 |
| frontal | supination | neutral | 13.0 | 6.5 |
| | pronation | neutral | 9.3 | 4.9 |
| sagittal | plantar flexion | 15° dorsal flexion | 57.5 | 21.3 |
| | plantar flexion | neutral | 50.6 | 22.0 |
| | plantar flexion | 15° plantar flexion | 36.3 | 16.3 |
| | plantar flexion | 30° plantar flexion | 20.5 | 11.5 |
| | dorsal flexion | 15° dorsal flexion | 35.4 | 19.2 |
| | dorsal flexion | neutral | 46.7 | 20.7 |
| | dorsal flexion | 15° plantar flexion | 50.2 | 20.3 |
| | dorsal flexion | 30° plantar flexion | 41.0 | 14.1 |

developed, based on Denavit–Hartenberg parameters, that led to the definition of a global transformation matrix by which the position and orientation of the foot are related to the shank. The theoretic study was completed by a definition of the Jacobian matrix of the mechanism, so that the last step of the characterisation could be performed: the determination of the joint kinematics and kinetics of the foot frame in the vectorial space described by the end plate of the mechanical chain.

The transducers and the electronic circuitry were also extensively characterised. The instrumental apparatus showed good linearity, precision and accuracy.

A preliminary, integrated experimental configuration was described and used for measurements on a healthy volunteer; it included the simultaneous use of a surface EMG amplifier and a multi-channel electric stimulator. The whole measurement system and the associated protocol showed adequate functional characteristics in terms of feasibility, reliability, robustness and repeatability. With regard to force moments and angular excursions acquired under voluntary and stimulated contractions of selected muscular compartments of the lower leg, a few interesting results are reported that highlight the main potentialities of the device

- to distinguish the moment components around the three anatomical axes, which is essential to discriminate between the contributions of the main muscles that act on the ankle articular complex
- to characterise muscles electrophysiologically in terms of the number of main motor points and their individual contribution to the global force the muscle exerts under different conditions of length and contraction
- to establish the stimulation frequency (25 Hz in the presented results) that best balances the need for significant contractions of the muscular fibres and the need for the correct EMG signal treatment with suitable filtering techniques.

The use of the device, together with a simplified measurement protocol on a sample of 20 healthy subjects, allowed us to establish reference values for angular excursions and moments. The analysis of flexion–extension moments under four different angular positions of the foot, with respect to the shank in the sagittal plane, highlighted the effect of muscle length on the produced moment. Interestingly enough, when the foot was 15° dorsiflexed, gastrocnemii gave higher values than in the neutral position, as expected (WINTER, 1992). It is worth noting here that, at 15° of dorsal flexion, the instantaneous axis of rotation of the ankle joint is forward with respect to the fixed, medio-lateral axis of the device (LEARDINI and O'CONNOR, 2002). Thus the measured plantar-flexion moment accounts for a longer lever arm.

For the dorsal–flexion moment, instead, the maximum value was obtained in relation to 15° of plantar flexion.

The real-time, visual feedback of EMG signals of the main involved muscles significantly helped in monitoring the initial conditions of muscle contraction and the correctness of the requested tasks. Several experimental studies, both *in vitro* and *in vivo*, and models have been discussed previously in the attempt to gain deeper knowledge of the physiology of the ankle complex. Lever arms of the main flexor and extensor muscles have been studied in the sagittal plane while the mechanical effects of the retinacula that constrain the tendons at the ankle joint were taken into account. Various inclinations of ankle and subtalar axes of rotation have been calculated. Some critical issues are still in question, such as the zero position of the subtalar joint and the movement of the whole ankle complex in the three reference planes under the effect of load.

The measurement system proposed in this study is based on a fixed reference system that hardly coincides with the instantaneous anatomical reference system. This constraint could represent a limitation in the accuracy of the measurements, but at the same time it delivers accurate, overall information on the simultaneous effect of

(i) changing positions of the rotation axis
(ii) the spanning of the tendons over several joints
(iii) the action of retinacula, sheaths or bony structures influencing muscle and tendon courses (LEARDINI and O'CONNOR, 2002).

When used in conjunction with the developed geometrical and mechanical models, the device may contribute to better knowledge of the biomechanics of the ankle complex.

From a clinical point of view, the device showed potential both as a diagnostic and a therapeutic tool and could be useful in monitoring the effects of conservative or surgical treatments. For this purpose, it will be significantly re-designed and implemented with suitable visual biofeedback techniques.

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All the authors are Researchers or Technicians at the Biomedical Engineering Laboratory of the Italian National Institute of Health. They are involved in designing and constructing mechanical and electronic devices for the analysis of human movement, and in clinical applications of dedicated tools and methodologies for diagnosis and therapy. As a Research Director, V. MACELLARI (Dr. in Electronic Engineering) was the main coordinator of the study and gave a contribution to each aspect of the project described in the paper. C. GIACOMOZZI (Dr. in Electronic Engineering, Ph.D. in Bioengineering), S. CESINARO (Dr. in Electronic Engineering), D. GIANSANTI (Dr. in Electronic Engineering, Ph.D.), G. MACCIONI (Technician) and M. PAOLIZZI (Dr. in Electronic Engineering) mainly dealt with the electronic issues described in the paper, while F. BASILE (Dr. in Mechanical Engineering), G. DE ANGELIS (Technician), E. MASCI (Dr. in Mechanical Engineering), A. PANELLA (Dr. in Mechanical Engineering, Ph.D. student), M. TORRE (Dr. in Mechanical Engineering) and P. VALENTINI (Dr. in Mechanical Engineering) mainly dealt with mechanical issues. C. GIACOMOZZI, S. CESINARO and A. PANELLA were also responsible for the experimental design and measurements, and for data elaboration.