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***Abstract* — Biomechanical analysis of Paralympic amputee sprinters has implications for the design of prosthetic components and for the improvement of the athletic performance. Field measurements are the optimal condition to collect data for this purpose. Therefore, the aim of this study was to complete a kinematic and kinetic analysis of a unilateral transfemoral amputee sprinter running on track. Data were gathered with a 28-camera motion capture (mocap) system and wearable kinetic sensors. Joint kinematics of the prosthetic limb and the intact limb were compared. The results presented here provide an example of outcomes which can support the improvement of** athletes' safety and performance.

***Keywords*—Running Biomechanics, Paralympic Sprinters, Load Cell, Motion Capture.**

I. Introduction

In the last decades there has been growing interest in the running biomechanics of individuals with lower limb amputation [1]. Research has allowed to gain greater insight into how lower-extremity amputees regain running capacity. Running Prosthetic Foot (RPF), carbon fibre blade with a spring-like function, has been introduced since (and a K over the years its design has been improved in order to better compensate for the replacement of an active leg.

It has been observed that laboratory measurements with subjects running on a treadmill, despite having high precision and resolution, cannot be entirely realistic, because sprinters are not used to running in such an artificial environment and at a constant speed [2]. In contrast, overground running tests can reproduce sprints very close to those performed during races. Thus, the results obtained from these tests better correlate with the actual athletic performance and are more useful for the structural optimization of running prosthesis. However, data collected in indoor or outdoor tracks are challenging for the cost of instrumentation, which should cover kinematics and kinetics over a convenient portion of the track.

Several studies make use of force platforms placed on the running track to measure Ground Reaction Forces (GRFs) during overground running [3], [4]. Except in cases where the number of platforms is very large [5], the results of this type of studies depend to a large extent on the probability of obtaining valid steps for the analysis (athlete hitting the plate and whole foot placed on the plate). Instrumented prosthesis have been demonstrated to overcome this issue by allowing realistic and numerous performance data to be obtained [6]. Various attempts have been made at kinematic analysis using inertial sensors atta (((less accurate than those obtained with the mocap system, considered the gold standard [7].

The purpose of the present study was to present kinematic and kinetic in-field collected data, obtained through an integrated motion capture and kinetic acquisition system, of a world-class athlete running unrestrained on a familiar track surface. These data will form the basis for a more in-depth analysis of the effects of the prosthesis configuration on running strategy and the minimization of injury risk.

II. Materials and Methods

A. Participants

One elite, gold medallist, Paralympic female athlete (mass: 55 kg; height: 1.60 m) with left unilateral transfemoral amputation, participated in this study. She wore the Ottobock 1E91 standard RPF, category 3.5, and she used the mono-axial Ottobock 3S80 prosthetic knee.

B. Instrumentation

Tests were performed in a 60m indoor athletics track (Palaindoor, Padova). In detail, the athlete ran on the eighth lane of the central straight runway.

Motion capture cameras (25 *Vero* (fs = 300 Hz) and 3 *Vantage* (fs = 300 Hz) - Vicon Motion System, Oxford, UK) were placed in the central part of the runway, covering an approximated 20-metres-long acquisition volume. 22 cameras were mounted on tripods placed along the runway, while the remaining 6 were installed on a double portal structure made of aluminium trusses with an overall size of 2.5 x 3 x 3.5 metres (length x width x height) positioned in the midpoint of the runway (Figure 1).

The Running-Specific Prosthesis (RSP) was instrumented with a 6-axis ultra-thin load cell mounted in between the prosthetic knee and the clamp of the RPF, and connected through wires to an ultra-small, standalone acquisition system (*DTS SLICE NANO* (fs= 2000 Hz) - DTS, Seal Beach, USA). The acquisition system was placed inside a small running backpack, together with its battery and a power bank for the load cell power supply.

C. Experimental protocol

The subject wore the prosthesis and the orthopaedic technician set the inclination of the socket with respect to the vertical line passing through the greater trochanter and the centre of the prosthetic knee. Specifically, the alignment adopted in the tests consisted in a socket pre-flexion angle (equal to 15°[8]. The athlete then warmed up to familiarize with the track and the prosthesis setup.

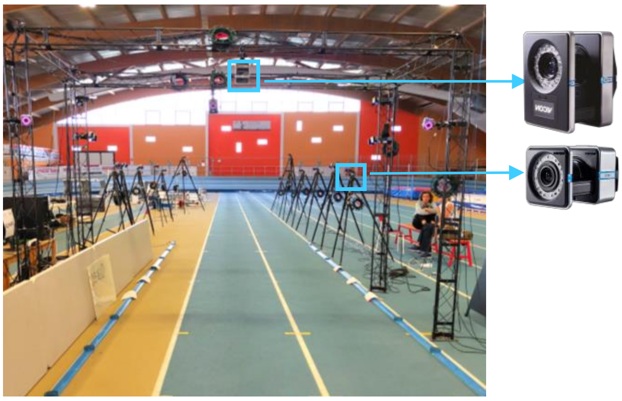

Fig. 1: Motion Capture system installed in the indoor athletics track. 28 Optoelectronic cameras (Vicon, fs = 300 Hz) placed on tripods and on the double portal structure.

Sixty-five retroreflective markers (14 mm diameter) were attached to the subject at specific anatomical landmarks and locations of the prosthetic components (Figure 2a, 2b), such as on the proximal and distal tip of the RPF, on the clamp, on the mechanical knee and on the socket. A wand calibration procedure was employed to identify 12 additional points, including four reference points on the loadcell, relative to technical clusters of markers. After the static acquisition, the calibrated markers, both placed medially on body segments (medial femoral epicondyle, medial malleolus, first metatarsal head, medial side of the prosthetic knee) and on the neck, thorax and posterior iliac spines were removed, because the former would have been likely to fall during the subsequent running trials and the latter would have been obstructed by the backpack and its straps.

The tests consisted of a series of sprints where the athlete started running from a line outside of the acquisition volume, accelerated to approximately 60% of her maximal speed (self-reported) and maintained this constant speed along the acquisition volume located 25 m from the start line. To avoid fatigue, the participant performed only three Steady State Running on track tests (TSSR).

D. Data analysis

Kinematic data were collected at 300 Hz. The labelling procedure was performed using Vicon Nexus 2.12 through an automatic algorithm. Subsequently, labelling was revised through careful frame-by-frame manual inspection to correct possible label swapping and gaps in the trajectories. Data were then low-pass filtered using a zero-lag fourth-order Butterworth filter with cut-off frequency equal to 10 Hz. Kinematic analysis was then carried out in MATLAB, with each body segment identified by a coordinate system (CS), i.e. an orthogonal triad defined through markers relative to that segment. The x-axis of the CS represented the anterior-posterior direction (pointing forward), y-axis the longitudinal direction (pointing upward) and the z-axis the medial-lateral direction (pointing rightward).

Ankle, knee and hip joint kinematics, as well as the prosthetic knee joint kinematics, were described by the orientation of the distal CS with regard to proximal CS using

instead. Trunk inclination in the sagittal plane was calculated as the angle with respect to the vertical of the vector joining

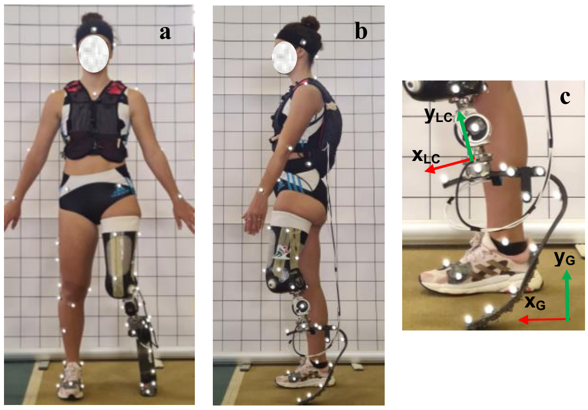

Fig. 2: 2a-b) Markerset adopted during the experimental session. Markers were applied on a headband worn by the subject, on arms, trunk, pelvis and right leg bony prominences and on the prosthetic components (socket, prosthetic knee, clamp and RPF). 2c) Detail of Load Cell coordinate system and ground coordinate system (sagittal view).

the centroid of the markers placed on the posterior iliac spines and the marker placed on the seventh cervical vertebra. The inclination in the sagittal plane of the unaffected shank (Theta Shank) and of the load cell installed in the prosthesis (Theta Load Cell) were likewise calculated as the angle between the longitudinal axis of the segment and the vertical global axis. As far as shank is concerned, longitudinal axis corresponded to the anatomical axis, while for the load cell it was the normal to the sensor surface.

Reaction Forces at the load cell were collected at 2000 Hz, therefore they were downsampled to 300 Hz and synchronized to correspond to kinematic data. Next, kinetic data were filtered using a zero-lag fourth order low-pass Butterworth filter with cut-off frequency equal to 52 Hz, which was the optimal value chosen from the spectral analysis. Eventually, vertical and horizontal forces expressed in the reference system of the load cell (and were resolved in the Ground reference system (and , by using the instantaneous orientation matrix of the load cell (Figure 2c).

Gait cycle events for the left leg were detected from filtered vertical force data, using a 5 N threshold, whereas for the right leg they were detected from the trajectory of the marker placed on the second metatarsal head, by identification of its minimum regions. Kinematic and kinetic data of each test were segmented into steps, averaged and normalized over the time of the right and left stride respectively.

Forward velocity was calculated as the distance divided by the elapsed time that the centroid of the pelvic cluster passed across the acquisition volume.

III. RESULTS

The results here provided are relative to the third test performed. Average speed over strides during this trial was 6.58 ± 1.12 m/s. Kinematic data are referred to both limbs, while kinetic data to the affected limb (AL) only, because in the unaffected limb (UL) neither force nor pressure sensors were used, and force platforms were not available.

A. Kinematics

The mean and standard deviation of lower-limb joint angles in the sagittal plane over stride time (5 strides AL, 4 strides UL) are reported in Figure 3, and analogous graphics were obtained for joint kinematics in the transverse and frontal plane.

During stance phase (0-25% of stride time) UL hip extended from 40° until reaching maximum extension of -10° at toe off. During swing phase it flexed up to 90° and then extended rapidly in the terminal part of swing. AL hip behaved similarly to the UL hip, but at foot contact it was 55° flexed, then it extended until foot off, reaching 12° as maximum extension angle.

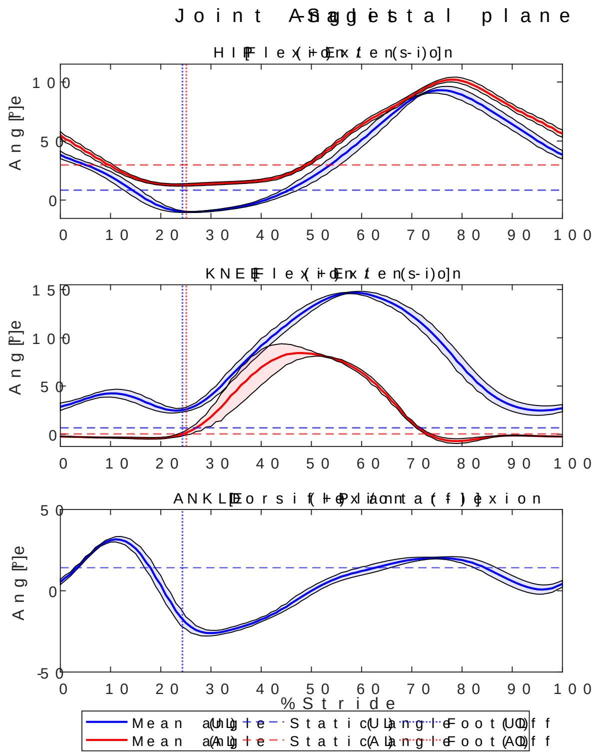

Fig. 3: Joint angles in the sagittal plane for hip, knee and ankle joint. Unaffected limb angles (mean 1 SD) are represented in blue and affected limb angles in red (mean 1 SD). Dashed horizontal lines represent the static angle (blue for UL, red for AL) and dotted vertical lines represent the average foot off instants (blue for UL, red for AL).

UL knee showed two periods of flexion, one during support and one during swing. Conversely, AL knee was completely extended during stance. The maximum flexion angle of the UL knee was nearly 60° higher than that of the prosthetic knee and it also happened later in swing.

The ankle joint was analysed for the UL only. Ankle was about 5° dorsiflexed upon foot contact, and then continued dorsiflexing rapidly to 31°. After midstance it plantarflexed and reached a maximum extension of -26° immediately after toe off. During the swing phase, ankle again dorsiflexed to 20°, and slightly extended prior to the next foot contact.

Figure 4a shows the trunk absolute inclination (positive forward). During AL support time the subject leaned forward, while during UL support time the trunk reduced its inclination. The total oscillation ranged from 3° to 18 °.

Figure 4b shows and , i.e. the inclination of the shank (UL) and of the load cell (AL) normalized to the AL stride time. Despite the offset, the two angles had a similar trend and similar ranges. Both angles had an average excursion of 50° from foot contact to foot off, but while (AL) increased from -25° to +25°, (UL) increased from +5° to +55°.

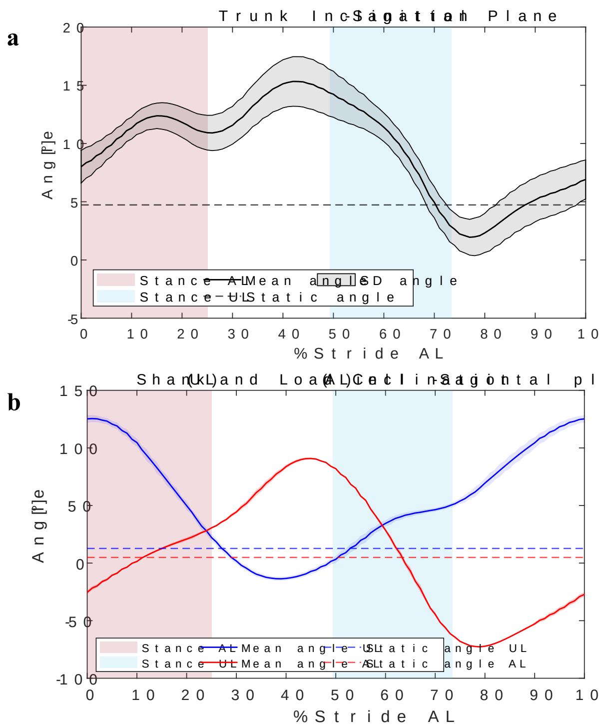

Fig. 4: Absolute angles in the sagittal plane of trunk (4a), shank and loadcell (4b) averaged and normalized over left stride time. Solid lines are the average angle during the running trial, horizontal dashed lines are the segments inclination in the static acquisition (black: trunk, blue: shank UL; red: load cell AL). Black shade in Fig. 4a is ±1SD relative to trunk average inclination. Red and blue vertical stripes represent the left and right stance time respectively.

B. *Kinetics*

Analysis of Reaction Forces (Figure 5) gave in output the average vertical and horizontal forces either in the Load Cell and in the Ground reference system. Vertical Forces, and , showed no relevant differences. They both peaked at nearly 3.5 N/BW, but in the second portion of the stance (40%-100%) decreased more rapidly than . While horizontal Force was always positive, was negative until 50% of stance and then became positive. Just before the horizontal GRF changed from braking to propulsive, turned from negative to positive.

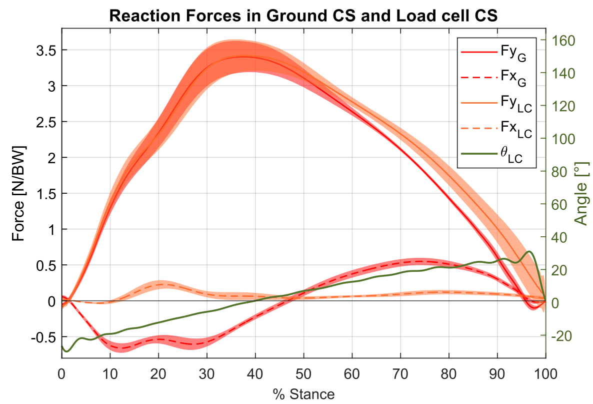

Fig. 5: Reaction forces measured by load cell averaged over 5 AL steps. Red lines represent the vertical (solid) and horizontal (dashed) forces in the Load Cell reference system (mean 1 SD). Orange lines depict the same forces projected onto the Ground reference system (mean 1 SD). Force values can be read in the left y axis, while the load cell inclination angle (solid green) can be read in the right y-axis.

IV. DISCUSSION

With regard to sprinting kinematics, in literature there is little information on the joint kinematics of para-athletes with unilateral transfemoral amputation. The present work showed that AL hip is expectedly more flexed than UL hip during the entire stride: this is related to the alignment of the prosthesis adopted, which allowed AL hip to work in a more comfortable range [8]. Substantial differences can be highlighted by comparing the joint angles of UL and AL knee, as the former is a biological knee, while the latter is a mechanical knee. In fact, due to its intrinsic functioning principle, AL knee did not flex during the stance phase, whereas UL knee did.

Many studies in the literature report the asymmetry between the GRFs exerted by AL and UL, pointing out that the GRFs in AL are lower than that of UL [3],[4],[6],[9]. Although inter-limb GRFs comparisons cannot be made in the present work, as no information on forces on the UL is available, we can observe that vertical peak force for the AL approached 3.5 N/BW, which is consistent with AL values presented elsewhere [6]. However, these values are much higher than the AL peaks reported in previous studies [3],[4],[9], but differences could be ascribed to inter-subject variability and the higher running speed reached by the athlete during the present pilot test. Kinetic results also differ from those obtained during tests with the same athlete running on a treadmill. This could be due not only to the fact that the athlete ran at a higher running speed (6.6 m/s vs 5 m/s), but also to the better confidence of running on track compared to running on the treadmill [10].

In the next months, the athletics track used in the present pilot test will be equipped with nine force platforms installed on the sprint track and on the long jump track, and with ten motion capture cameras hanging from a moving support structure [11]. This configuration will allow for a more extensive data collection and subsequent comparison of multiple subjects.

V. CONCLUSIONS

This paper illustrated an experimental procedure to collect in vivo-indoor data during a pilot test of an elite Paralympic athlete running on a track. Mocap cameras and a wearable kinetic sensor installed in the RPF were used. Biomechanical analysis revealed differences between AL and UL joint kinematics and revealed high GRF values. These results could be useful for athletes and trainers in the fine-tuning of running strategy. However, it should be noted that this is a single case study, so the results will have to be confirmed by a larger analysis including more trials and more subjects, which will be carried out in the forthcoming months.

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