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Planar covariation of elevation angles in prosthetic gait

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ABSTRACT

In order to achieve efficacious walking, transfemoral amputees must adapt coordination within both the artificial and the sound lower limb. We analyzed kinematic strategies in amputees using the planar covariation of lower limb segments approach. When the elevation angles of the thigh, shank and foot are plotted one versus the others, they describe a regular loop which lies close to a plane in normal adults' gait. Orientation of this plane changes with increased speed, in relation to mechanical energetic saving. We used an opto-electronic device to record the elevation angles of both limbs' segments of novice and expert transfemoral amputees and compared them to those of control subjects. The statistical structure underlying the distribution of these angles was described by principal component analysis and Fourier transform. The typical elliptic loop was preserved in prosthetic walking, in both limbs in both novice and expert transfemoral amputees. This reflects a specific control over the thigh elevation angle taking into account knowledge of the other elevation angles throughout the gait cycle. The best-fitting plane of faster trials rotates around the long axis of the gait loop with respect to the plane of slower trials for control subjects, and even more for the sound limb of expert amputees. In contrast, plane rotation is very weak or absent for the prosthetic limb. We suggest that these results reveal a centrally commanded compensation strategy.

1. Introduction

Numerous experimental studies of locomotion have demonstrated that the elevation angles of the lower limb segments provide a kinematic template revealing a motor organizational rule [1–6]. Intersegmental coordination of the lower limbs could be used by the nervous system for limiting energy expenditure to walk in a smooth and effortless manner [7]. When the elevation angles of the thigh, shank and foot are plotted one versus the others, they describe a regular loop which lies close to a plane. Lacquaniti et al. [3] calls this the "first law of intersegmental coordination". The orientation of this plane and the shape of the loop are similar for normal subjects [6]. Moreover, the orientation of the plane slightly changes with increased walking velocity, and this plane rotation limits the increment in mechanical energy expenditure; this is considered as the "second law of planar covariation" [3,7,8]. This physiological approach to gait control has proved useful to better understand a number of gait disorders associated with various conditions of the brain, whether acquired such as Parkinson's disease [9] or developmental, such as cerebral palsy or Angelman syndrome [10], or the spinal cord [11,12]. It has also been tested in situations where the peripheral system is artificially altered, such as walking on a slippery surface [13] or reducing joint mobility [14]. Here we studied the reorganization of lower limb coordination in patients who developed prosthetic gait in the context of transfemoral amputation to see whether a transfemoral amputee will simulate the specific coordination observed in normal subjects. We analyze the prosthetic lower limb kinematics of transfemoral amputees at different stages of rehabilitation, walking at different speeds in order to verify the existence of both laws of elevation angle covariation. Amputees adjust to a device over time, essentially optimizing their physiological system with that of the prosthesis [15]. Thus, we have to consider the amputee's physiological system not as a fixed component but as a compliant biological one like that observed in typical gait maturation [16]. As such, we also see how the compensatory strategies of unilateral amputees affect their sound limb's coordination pattern. We included the very first walking trials of fresh amputees, even bilaterally amputated, to check whether the first law of planar covariation could be found within a very heterogeneous group of amputees.

Only expert amputees, able to walk at different speed levels on a treadmill, allowed testing the existence of the second "law of planar covariation". The main purpose of this study is to highlight

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specific differences between a sound and a prosthetic limb in the walking strategies of expert amputees.

Additionally, the results of this study might give insights into the physiological relevance of the first and second law of planar covariation. A key issue concerns the role of the thigh elevation angle, which could either be marginal with respect to the correlation between shank and foot elevation angles [17] or contribute independently to the pattern of intersegmental coordination and planar covariance. Analyzing the prosthetic limb of an amputee, where knee and ankle joint functioning are based on a completely passive mechanism and on the voluntary forward and backward oscillation of the thigh, might allow gaining further insight to these questions.

2. Methods

A total of 20 subjects participated in this study (males, age: 35 ± 6 years): seven expert transfemoral amputees walked as naturally as possible on a treadmill at various self-selected speed levels, reaching their self selected maximal speed; two bilateral transfemoral amputees walked on the ground with two crouches after only two months of rehabilitation and one unilateral amputee, with his recently matched prosthesis performed his first steps without crouches in the laboratory setting.

The expert amputees were used to walking with a conventional hydraulic or pneumatic knee and a conventional prosthetic foot, without energy-storage and return. The more recent amputees all wear microprocessor-controlled prosthetic knees and conventional feet (C-Leg, Otto-Bock, SACH foot). All have a relatively long stump without any movement restrictions or skin problems.

Ten aged-matched healthy adults without any gait, neurological or orthopaedic impairment served as controls. They walked at the same speed levels on the treadmill as the amputees and faster, reaching their self selected maximal speed.

Kinematic data was recorded at 100 Hz by means of a 6-camera opto-electronic ELITE system (BTS, Milan, Italy). This system recognizes multiple passive markers fastened onto the skin overlying bony landmarks of the sound limb and over the corresponding points of the prosthetic limb (Fig. 1).

For treadmill walking, all subjects performed a training session, where they walked at different self-selected speeds on the treadmill in order to get used to treadmill walking.

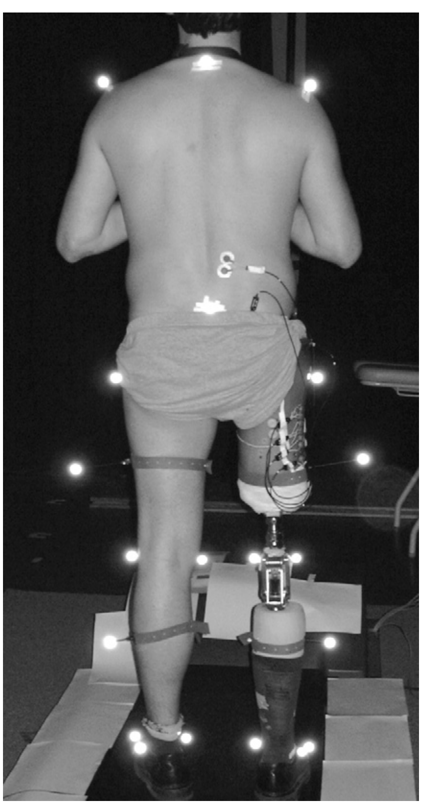

Fig. 1. Photograph of one of the expert amputees during standing acquisition on the treadmill. The picture also shows the specific shaft that had been developed with the purpose to acquire EMG of stump muscles using integrated electrodes.

The elevation angle of a limb segment is defined as the orientation of the segment with respect to the vertical and the walking direction and is positive in the forward direction. Thigh, shank and foot elevation angles in the sagittal plane (α_T, α_S and α_F, respectively) (Fig. 2) are smoothed by a low-pass filter with a cut-off frequency of 6 Hz.

The methods for analyzing the planar covariation of elevation angles were the same as those used by Borghese et al. [6]. The statistical structure underlying the distribution of geometrical configurations associated with the observed changes of the elevation angles were described by principal component (PC) analysis. The PCs were computed for five gait cycles after subtraction of the mean value, and identified the best-fitting plane of angular covariation for each speed and each limb. The residual percentage of variance accounted for by the third and last PC, is an index of planarity of the loop (0% corresponds to an ideal plane). For each trial, the eigenvectors of the covariation matrix of the ensemble of time-varying angles were computed. The first two eigenvectors U1 and U2 lie on the best-fitting plane; while the third eigenvector called U3 is normal to the plane and defines the orientation of the plane in the 3-D position space. The direction cosines of the plane normal U3 with the semi-positive axis of the thigh, shank and foot angular coordinates are called U3T, U3S and U3F, respectively (Fig. 2). In order to verify the origin of the planarity and the orientation of the covariation plan in amputees, we used the methods proposed by Barliya et al. [18]. As the angular profiles of all three segments are close to sinusoidal oscillations, the time shift between angular motion profiles can be assessed by means of a Fourier decomposition of the thigh, the shank and the foot elevation angle. The basic frequency, the amplitude and phase of the first harmonic of these three angles versus walking velocity will be computed.

Statistical analysis is performed using Statistica software (StatSoft Inc., Tulsa, USA), ANOVA testing allowed the comparisons between the parameters of the planar covariation between the sound and prosthetic limb.

All subjects signed the informed consent according to the Helsinki's declaration and the procedures have been approved by the local ethics committee of the University.

3. Results

Evolution of elevation angles of the three segments of the lower limb was studied throughout gait cycles. Fig. 2 shows this evolution in one representative amputee (left and middle column for prosthetic and sound leg, respectively) and in one representative control subject (right column) all walking at 3 km/h. When plotting the time-varying series of elevation angles shown in Fig. 2A one versus the others in 3D-position space, we found a planar covariation for both the prosthetic and the sound limb (Fig. 2B). The typical elliptic shape found in normal gait looks like a footprint with a marked 'big toe' and is also found in the prosthetic leg. The extremity of the "big toe" corresponds to heel strike, the points of the three angles are following the ellipse in counterclockwise direction, the lower extremity of the ellipse representing toe off. Fig. 2C shows a profile view of these covariation planes. In the left column, representing the prosthetic side, the points deviate from the plane only during the first part of stance phase, before the heel lifts off. In contrast, for the sound limb (middle panel) and for the control subject (right panel), the lower tip of the loop also slightly deviates from the plane. Overall, more than 98% of the total experimental variance was accounted for by the first two PCs. Even for novice and for bilateral amputees planarity was extremely well observed. If we compare the residual variance (PV3) found in all prosthetic limb data ($n = 51$, as we consider all speed levels in expert amputees and add the data from novice amputees) with PV3 for sound limbs ($n = 47$, as there are two bilateral amputee), we find by a one-way ANOVA that PV3 is significantly smaller for the prosthetic limb (0.32 ± 0.13) than for the sound limb (0.59 ± 0.21) ($F(1, 96) = 60.545$, $p = .00000$).

The values of U3T associated with walking speed for the prosthetic side (Fig. 3, crosses) were compared to the sound limb (Fig. 3, triangles). This is represented for each expert amputee (designated by SA1–SA7) against the distribution of the experimental data recorded in the control group (Fig. 3, circles and shaded area). Despite individual variation in the control group U3T tends to decrease linearly with speed beyond a threshold. This threshold is subject-dependent and varies from 1.00 to 1.2 m/s in control subjects. In contrast, for the sound limb of the amputees,

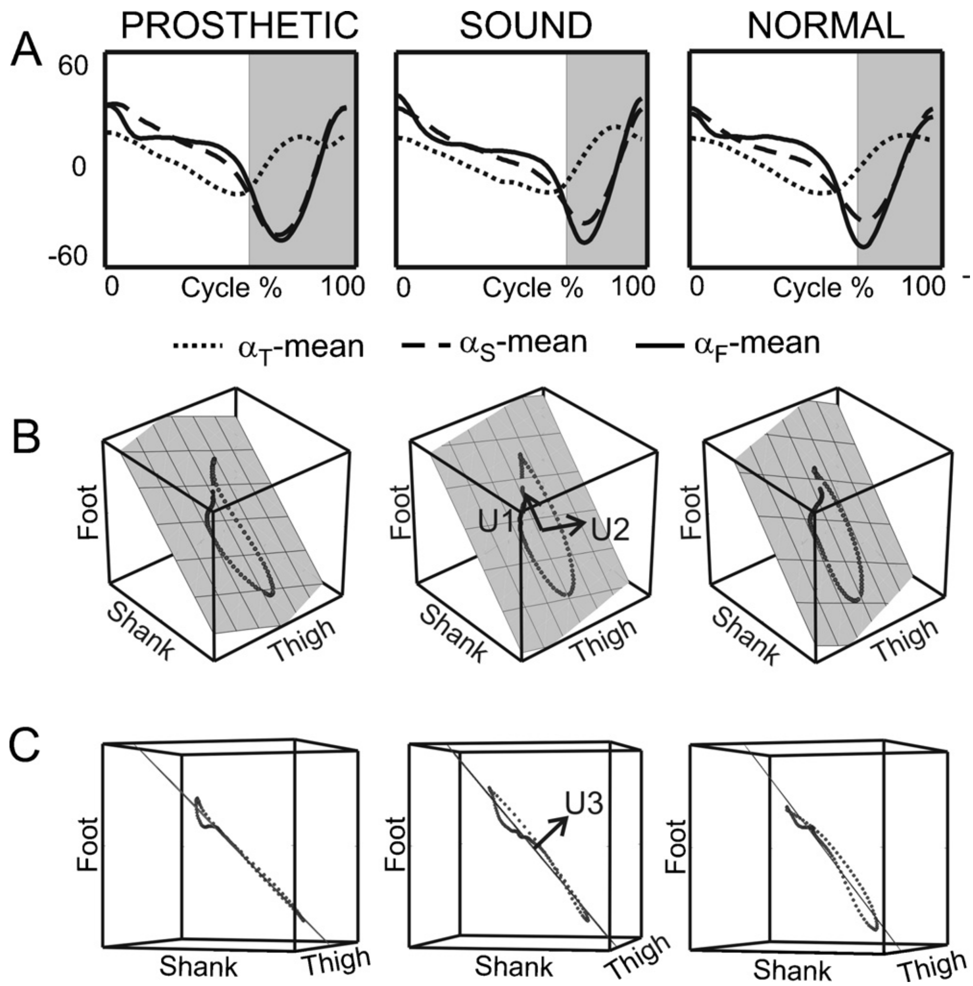

Fig. 2. (A) Mean elevation angles of the thigh (dotted line), shank (interrupted line) and foot (continuous line) of one representative amputee walking at 3 km/h, assessed respectively from the prosthetic side (first column) and the sound side (middle column), compared to those of one representative control subject walking at the same speed (last column). The grey shaded part of each graph corresponds to the swing phase. Frontal (B) and profile (C) view of planar covariation of the elevation angles shown in (A) during one gait cycle. The grid corresponds to the best fitting plane.

U3T decreases with speed for much slower walking velocities, often even, from the first speed levels around 0.5 m/s. Simple linear regression between U3T and speed was computed for each subject and limb. U3T is related to speed as shown by a p-value below 0.05 for all sound limbs, but only for four of the seven prosthetic limbs. The mean slope value for these subjects decreases from -0.20 ± 0.08 for the sound limb to -0.07 ± 0.04 for the prosthetic limb.

In order to better explain the origin of the planarity and plane orientation Fourier transform was applied to the elevation angles. We thus found that the first harmonics of the Fourier series of the thigh, shank and foot elevation angles account for the greatest part of the total variance of the first 10 harmonics in control subjects, prosthetic and sound limb of amputees. According to the mathematical model of Barliya et al. [18], the frequency, phase and amplitude of the first harmonic is sufficient to explain planarity and plane orientation in healthy subjects. We therefore limited further analysis on the characteristics of this first harmonic. The fundamental frequencies of all segments of a limb were equal for a given speed, for both amputees and control subjects and were a function of speed ($r = 0.861$; $p = 0.0000$; $y = 0.523 + 0.304x$) (Fig. 4A). However, we observe very small differences between the sound and the prosthetic limb (Fig. 4A). These differences disappear if we express the basic frequency as a function of cycle duration ($r = -0.968$; $p = 0.0000$; $y = 1.572 - 0.599x$) (Fig. 4B). Indeed, per gait cycle, the first harmonic of each segment oscillates once forward and backward. Fig. 5A shows the shank–foot phase lead as a function of walking velocity in the control subjects' limbs (open circles), the sound limbs of the expert amputees (triangles) and the prosthetic limbs of the expert amputees (crosses). When results were pooled together, no consistent phase lead was found. However, computing the individual slope of the shank–foot phase lead versus speed demonstrates a significant relationship in all sound limbs, but no relationship in four out of seven prosthetic limbs, while for the remaining three limbs the mean slope value decreases from -12.89 ± 9.09 for the sound limb to -5.04 ± 2.81 for the prosthetic limb. However, when all expert amputees are taken together, no significant relationship between this phase lead and speed could be found for prosthetic limbs ($r = -0.272$; $p = 0.070$; $y = 8.706 - 2.747x$), but exists for sound limbs ($r = -0.312$; $p = 0.039$; $y = 14.801 - 7.066x$) and of course for control subjects ($r = -0.46$; $p = 0.0008$; $y = 18.7358 - 4.6309x$). The correlation between walking speed and U3T for all analyzed subjects has a similar appearance (Fig. 5B). Interestingly, a tight correlation between shank–foot phase lead and U3T ($r = 0.9242$; $p = 0.0000$; $y = -1.356 + 58.453x$) was found for all types of limbs considered together (Fig. 5C). This means that for amputees as for control subjects, plane rotation is depending on the shank–foot phase lead of the first harmonic of these elevation angles.

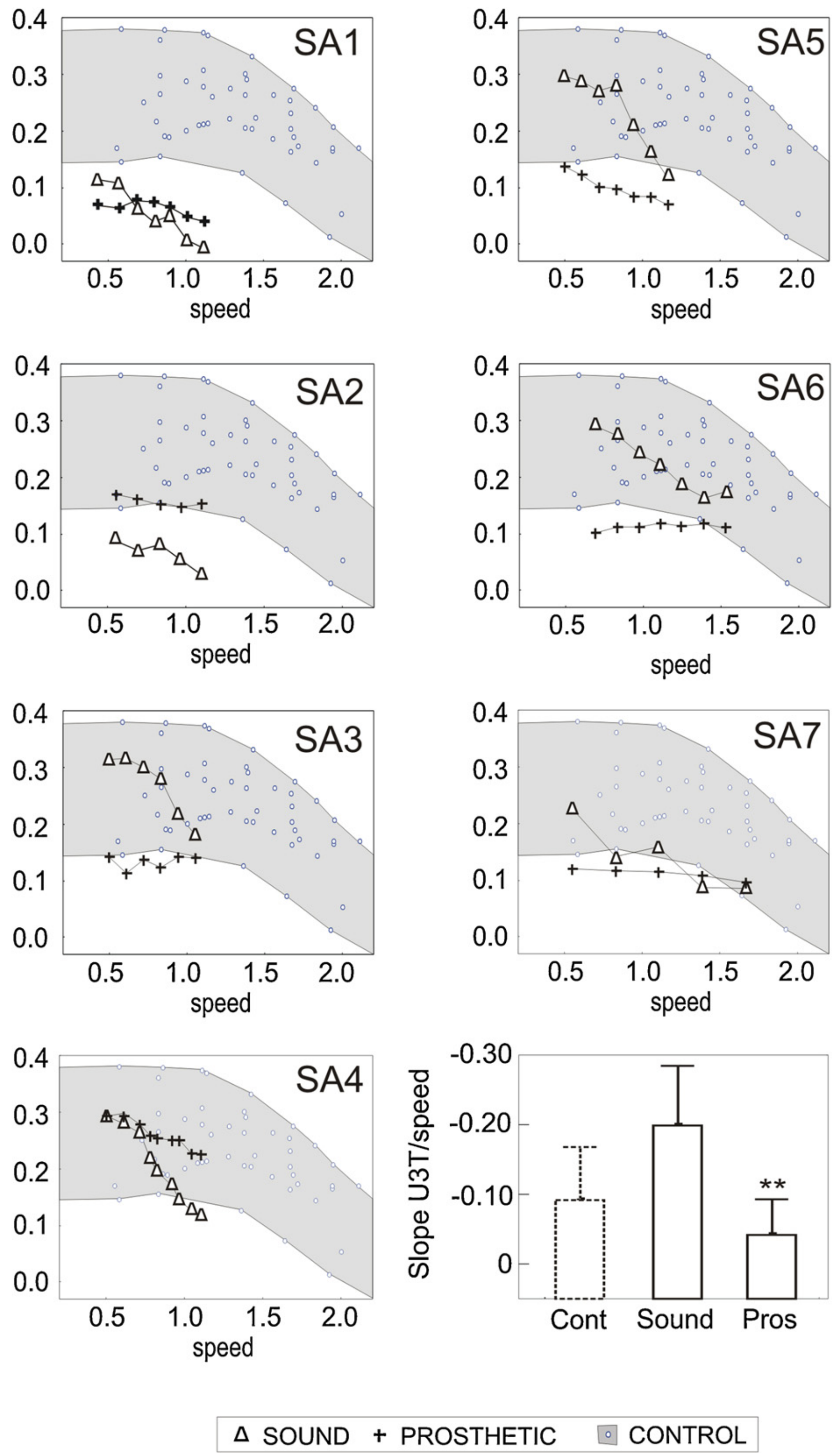

Fig. 3. Scatterplot of the direction cosines of the plane normal U3 with the semi-positive axis of the thigh (U3T) compared to walking velocity (speed in m/s). The values of U3T are represented by crosses for the prosthetic side and triangles for the sound limb individually for each expert amputee (designated by SA1, SA2, SA3, SA4, SA5, SA6 and SA7). The underlying grey surface delineates the values in the control group (open rounds). The histogram corresponds to the slope of UT3 versus speed in the control (Cont), sound (Sound) and prosthetic (Pros) limb, the asterisks mean statistical significance ($p < 0.01$) between the sound and prosthetic limb.

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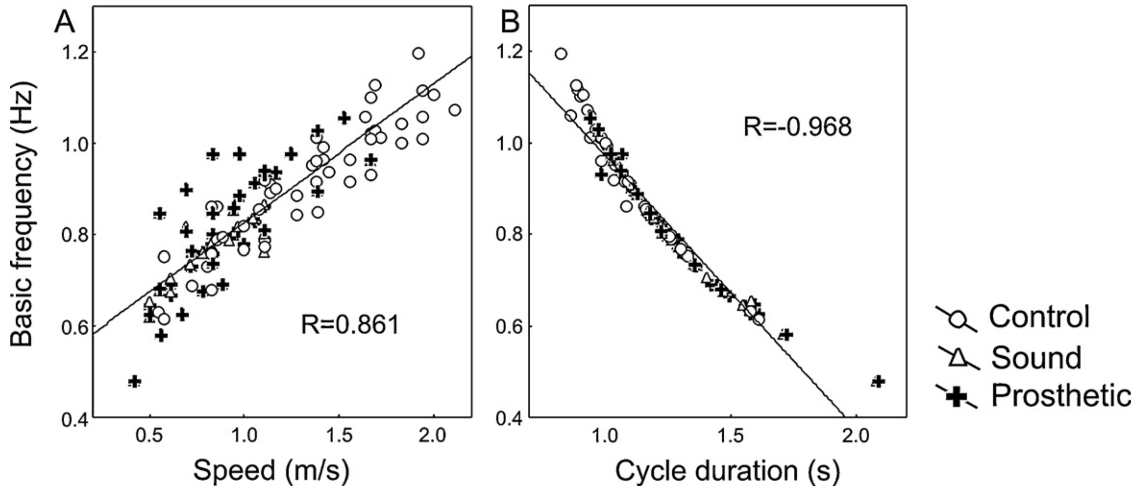

Fig. 4. (A) Basic frequency (in Hz) of the first harmonic of the three elevation angles as a function of speed (in m/s) respectively for control subjects (circles), the sound limbs (triangles) and the prosthetic limbs (crosses). (B) Same basic frequencies expressed as a function of cycle duration, which means time from one heel strike to the next of the same limb (expressed in s).

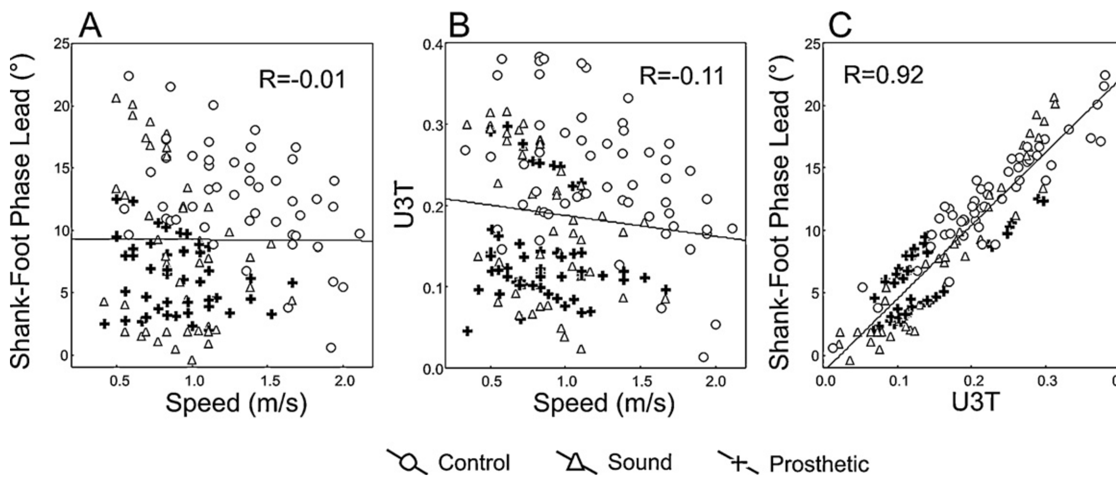

Fig. 5. (A) Scatterplot of shank–foot phase lead as a function of walking speed for all analyzed limbs taken together, namely the control's limbs (open circles), the sound limbs of the expert amputees (triangles) and the prosthetic limbs of the expert amputees (crosses). The continuous line represents the best fit for all subjects taken together while the '*r*' represents Pearson's correlation coefficient. (B) Same scatterplot for U3T compared to walking velocity. (C) Shank–foot phase lead as a function of U3T.

4. Discussion

We found that the first law of intersegmental coordination of elevation angles was verified in both limbs of transfemoral amputees, whether novice or expert in the use of their prosthesis. This contrasts with findings in the gait pattern of young toddlers, which differs significantly from the typical planar covariation that emerges over the first few months and years of walking experience in normal development [16,19–21]. In fact, in amputees, planarity of the covariation was greater for the prosthetic than for the sound limb.

This observation could be consistent with Hicheur et al.'s [17] suggestion that planarity of the thigh, shank and foot covariation could be explained by a linear relation between the shank and foot segments, without any contribution of the thigh angle. However, the observed pattern of covariation within the plane, namely the typical elliptic shape, which graphically looks like a foot-print with a marked 'big toe', is also found in prosthetic walking. This likely reflects a specific control over the thigh elevation angle taking into account knowledge of the state of the other elevation angles at each point in time during the gait cycle [18]. Ivanenko et al. [22] observed that a perfect correlation between the shank and foot implies that the plane of intersegmental coordination is parallel to the thigh axis, which is not the case here.

A recent mathematical model presented [18] offers a more complete explanation of planar covariation, by analyzing elevation angles as quasi-sinusoidal signals. In normal subjects walking overground, the basic frequencies of elevation angles of the three segments are roughly equal, which accounts for planarity. We made similar observations in both limbs of our amputees. In addition, we showed that the basic frequency is tightly correlated to cycle duration in all experimental situations.

Moreover, the phase delay between the prosthetic shank and foot is smaller than in the other cases, which may explain better planarity for the prosthetic side, as Barliya et al. [18], showed that planar covariation corresponds to equal phase shift between shank and foot segments.

Consistently with Bianchi et al. [7,8], we found that the best-fitting plane of faster trials rotates around the long axis of the gait loop with respect to the plane of slower trials for control subjects, but the comparison of the slope of U3T versus speed between control and amputated subjects is blurred by the fact that at slower speed levels U3T of control subjects remains stable before decreasing at higher speed levels. However we could demonstrate that plane rotation is much more important in the sound than in the prosthetic limb. This may indicate that future neuro-feedback systems should be developed from the sound limb kinematics in order to facilitate a fine tuned coordination and energetic savings in prosthetic walking.

According to Ivanenko et al. [23], the reason for plane rotation might not be the same for different gait types and thus cannot automatically be related to segment angle phase relationship, as in normal gait. Considering prosthetic gait as a specific gait type we show here that plane rotation is in tight relationship with the shank–foot phase lead for all analyzed gaits, although for the artificial limb plane orientation differs from speed-matched plane orientation in normal gait.

In the present study, only conventional prosthetic feet were analyzed, regulating the shank–foot motion in a completely passive way. We may thus reformulate Hicheur's suggestion that the first, but certainly not the second law of planar covariance may be an outcome of passive rather than active coupling between segment angles. Many authors suggested that elevation angles were a reflection of central constraints [3,9,11,24,25]. Here we show that mainly the phase relationship between segments is centrally controlled.

Van der Linden et al. [26] studying the effects of various types of prosthetic feet on the gait of transfemoral amputees show that analyzing the sound limb provides important information on the performance of prosthetic feet. We suggest that the faster evolution of shank–foot phase shifts with speed seen in the sound limb may be interpreted as a compensatory strategy at the level of the sound limb. Indeed, Hofstad et al. [27] showed that in unilateral amputation, a basic reorganization within the central nervous system concerns both lower limbs, even though the peripheral deficit involves only one limb. Laroche et al. [28] suggest that in gait impairment, sensory signals and pain help to update the motor command and compel patients to adjust the descending command to the altered limb representation. We suggest that bionic feet might allow reproducing the phase shift between shank and foot. However, most of the 'bionic' devices are still on the research level nowadays, but one can expect that they will become available on the market soon [29]. Studying the phase relationship between segments with varying speed could be a promising way of assessing efficiency in prosthetic gait and may help to design feedback controllers for brain–machine interface in rehabilitation [30].

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Conflict of interest statement

Concerning our manuscript entitled 'Planar covariation of elevation angles in prosthetic gait', all authors disclose any financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work Leurs F, Bengoetxea A, Cebolla AM, De Saedeleer C, Dan B, Cheron G.

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