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Angular displacements in the upper body of AK amputees during level walking

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Abstract
The angular displacements of the longitudinal axis of the trunk, and of the latero-lateral axes of pelvis and shoulder girdle were measured in five normal subjects and four AK amputees during level walking at different speeds. Amputees used single axis prostheses with the SACH foot. Spatial measurements were carried out in three dimensions by means of a photogrammetric technique. The time functions of the target angles underwent harmonic analysis. Based on the Fourier coefficients, comparison was made between normal subjects' and amputees' angular displacements. Relevant findings permitted the identification of compensatory mechanisms adopted by amputees at trunk level as well as the assessment of the relationship between these latter mechanisms and those put into action at lower limb level.

Introduction

The movement of the upper body constitutes an effective reference for the assessment of the quality of gait (Saunders et al. 1953; Murray et al. 1964; Lamoreux, 1971; Cappozzo et al. 1978a).

In this paper rotational displacements of the upper body during level walking of AK unilateral amputees and normal subjects are analyzed. The rotations taken into account were the frontal and horizontal rotations of the transverse axis of the shoulders and of the pelvis, and the frontal and sagittal rotations of the longitudinal axis of the trunk. This choice corresponds to a well established tradition in the field of biomechanics and refers to easily understandable kinematic quantities. The pelvic rotations have been classified as gait determinants by Saunders et al. (1953). The shoulder and trunk rotations are correlated with the maintenance of balance and with the mechanical energy efficiency of the locomotor act (Cappozzo et al. 1978a). The analysis of these rotations can provide useful data for both amputee's gait evaluation and improvement of prosthesis design, provided that it is carried out quantitatively and an easily readable synthetic description of the relevant time functions is devised.

Since walking is a cyclic movement, the related kinematic variables can be represented through a Fourier series. Each Fourier component is a sinusoid with a period equal to the stride period or to an integer submultiple of it; it is fully described by only two parameters: the amplitude and the phase. The harmonics associated with human locomotion are very few (Bernstein, 1966; Winter et al. 1975; Cappozzo et al. 1979a;), therefore the most relevant information relative to the investigated kinematic quantities is contained in a few numbers only.

Materials and methods

Five normal male subjects and four male amputees were tested during level walking at various speeds. A total of 29 tests were carried out. The subject anthropometric data and the main characteristics of the tests they were subjected to are shown in Table 1. During the walk trials all subjects wore their usual shoes. The experimental set-up has been completely

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Table 1. Subject and test data.

Group	Subject	Age (yr)	Height (m)	Weight (kg)	Amp. side	Amp. level	Test	Speed (km/h)	Stride period (s)
Normal subjects	1	22	1·72	70			1	4·49	1·15
							2	4·95	1·10
							3	5·94	1·03
							4	5·94	1·01
	2	23	1·73	70			1	4·55	1·20
							2	4·96	1·13
							3	5·23	1·10
	3	21	1·76	66			1	3·57	1·24
							2	5·56	1·05
	4	32	1·81	72			1	4·30	1·31
							2	4·78	1·20
							3	5·80	1·05
	5	22	1·86	66			1	5·05	1·10
							2	6·01	1·02
Amputees	6	18	1·67	56	L	I	1	2·80	1·43
							2	3·00	1·37
							3	3·60	1·14
	7	18	1·68	56	R	II	1	3·20	1·29
							2	3·50	1·27
							3	3·50	1·30
							4	3·90	1·31
							5	4·60	1·20
							6	4·70	1·10
							7	5·00	1·09
							8	5·20	1·03
	8	31	1·68	91	L	II	1	2·34	1·42
							2	2·40	1·45
							3	2·50	1·34
	9	37	1·74	74	L	II	1	3·60	1·25

described previously (Cappozzo et al. 1978; Cappozzo, 1981), however, it is important to note its stereophotogrammetric nature, and that it yields an overall precision in the assessment of any body point position better than 5 mm, to which an indetermination less than 0·5 degrees on the considered rotations corresponds. Markers were placed on the acromial process and tubercule of the iliac crest of both the right and left side. Their position was sampled at a sampling frequency (fs) equal to 30 Hz and 60 Hz for low (less than 5 km/h) and high speeds respectively.

Results

Each considered rotation was described as follows:

$$\theta(t) = M_o + \sum_{i=1}^{N} M_i \sin(i\Omega t + \Phi_i)$$

where:

M_i = amplitude of the ith harmonic (i = 1,...,N);
Φ_i = phase of the ith harmonic, with reference to a time origin coinciding with left heel strike;
Ω = fundamental frequency, corresponding to the stride period;
M_o = mean value of θ (t) over the stride period;
N = maximum order of the significant harmonics (never greater than 4).

As far as the resulting precision in calculated Fourier coefficients calculation (Cappozzo et al. must be stressed; precision is mainly affected by the eventual inexact correspondence of the stride period value to an integer multiple of the sampling period (maximum difference $\pm\frac{1}{2fs}$) and by the statistical characteristics of the residual measurement noise. The standard deviation of the latter was estimated by the procedure of Fourier coefficients calculation (Cappozzo et al. 1975) and was recognised to be never greater than 0·1 degrees. According to these figures the error on the amplitude and phase of each harmonic was estimated and resulted in values never greater than ± 0·5 degrees for the amplitude and ± 1 degrees for the phase.

An example of the quantities under study and the way they were represented is shown in Figure 1. Each harmonic is fully described by a vector in the polar plane. Figure 2 shows the effect of different harmonic phase values on the shape of the resultant $\theta(t)$ vs time plot. The harmonics that should be the only result of the Fourier analysis using the hypothesis of complete symmetry of the movements were termed intrinsic to the locomotor act, correspondingly the other harmonics were defined as extrinsic (Cappozzo et al. 1979a; Cappozzo, 1981). In accordance with elementary geometric considerations, the second and fourth harmonics are intrinsic for trunk sagittal rotation, while the first and third harmonics are intrinsic for all other rotations.

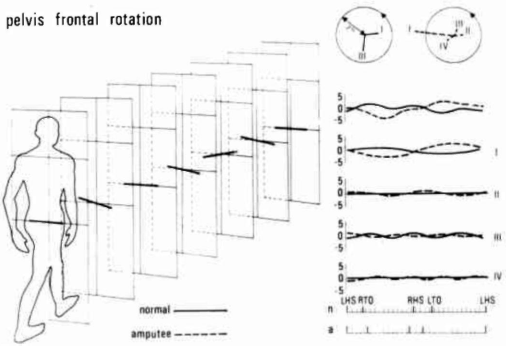

Fig. 1. Frontal rotation of the pelvis; the rotating element is the projection on the frontal plane of the segment. The angle vs time plot is represented together with the plots of its four harmonic components; at each time the value of the angle is the sum of the corresponding values of the harmonics plus the mean value (not represented). Each harmonic is described by the vector having its amplitude and its phase; the corresponding time law can be obtained by plotting the projection on the vertical axis of the actual position of the vector, provided it rotates with a constant velocity, equal to $i\Omega$, and starts from the plotted position at $t = 0$.

In Figures 3, 5 and 6 the envelope diagrams of the vectorial representation of the harmonics in the polar plane are given for all the tests performed on the normal as well as for the amputee subjects.

It is important to stress that, in general, the large scatter of phase and amplitude values shown by some intrinsic harmonics of the normal subjects cannot be correlated with the speed of progression. In general the harmonics of the same subject show good repeatability; the scatter of the diagrams is mainly due to the inter-

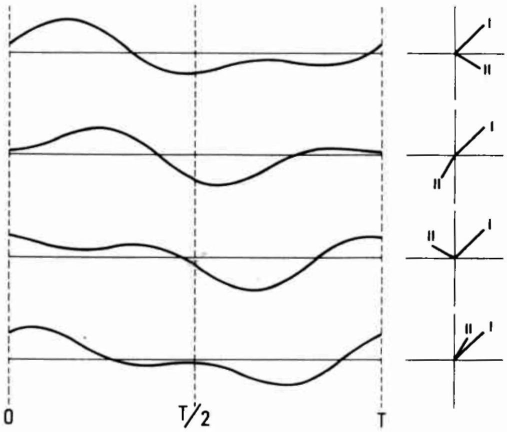

Fig. 2. The relationships between the phase values of the harmonics and the shape of the resultant functions are shown. The shapes of four functions resulting from two harmonics (first and second) equal in all except the phase of the second one is presented as a didactic example. Near each plots vs time the corresponding synthetic representation by means of vectors in the polar plane is shown.

subject differences. The same can be said with respect to the large scatter of the amputee's intrinsic harmonics.

Pelvic rotations (Fig. 3, 4)

1) Normal subjects

Both the frontal and horizontal rotations were found to be consistent with the findings of Murray et al. (1964) and Saunders et al. (1953). These rotations were remarkably well defined by the (intrinsic) first and third harmonics only.

With reference to the *frontal plane* the following was noted; (a) the first (intrinsic) harmonic covered a limited sector of the polar plane, ranging in phase from $-15°$ to $70°$ and amplitude values centred at approximately $1{\cdot}5°$; (b) the third harmonic also covered a limited sector of the polar plane with the phase centred on $-95°$ and the amplitude ranging from $0{\cdot}7$ to $1{\cdot}8°$; (c) the second and fourth (extrinsic) harmonics were minimal and could, thus, be disregarded.

Regarding the *horizontal plane;* (a) the first harmonic amplitude was relatively large (from $1{\cdot}4°$ to $4°$), its phase was widely scattered, but the repeatability within the same subject was good, as shown in Figure 4. The same characteristics were shown by all the intrinsic harmonics of the

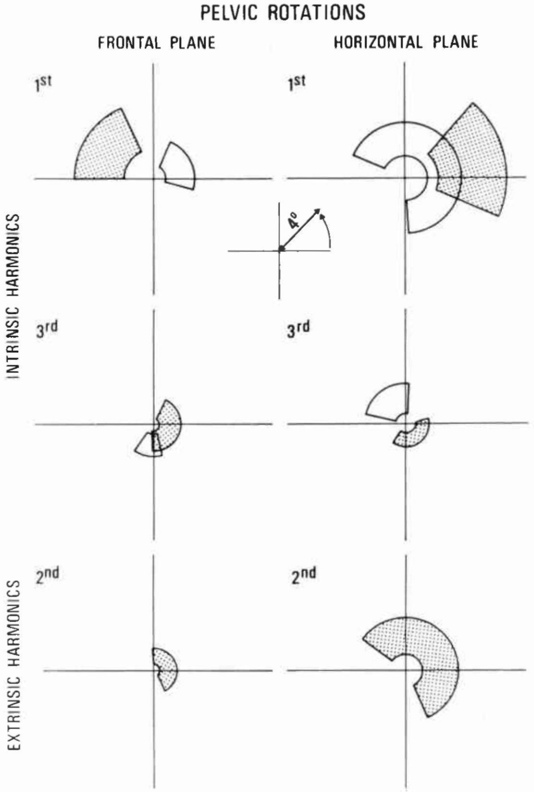

Fig. 3. Synthetic representation of the pelvic horizontal and frontal rotations in normal subjects and amputees; intrinsic and extrinsic harmonics are separately drawn. The field covered by amputee tests is represented by shaded sectors.

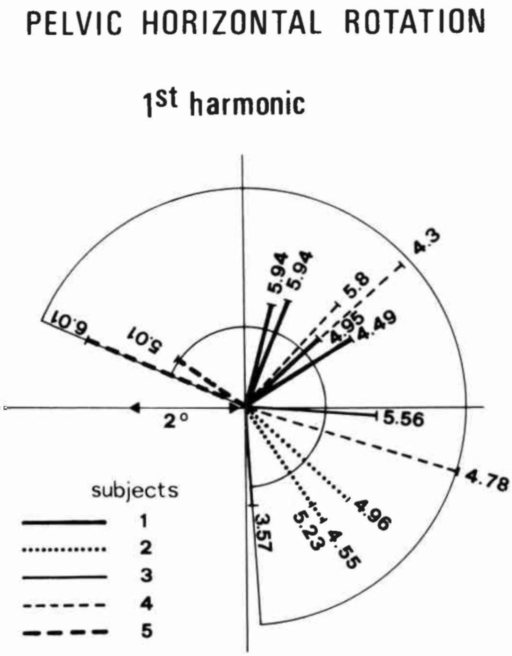

Fig. 4. First harmonic of the hip horizontal rotation in normal subjects. Each test is labelled with the corresponding speed of progression (km h^{-1}).

different rotations taken into account in this paper;

(b) the third harmonic was less scattered in phase and the amplitude ranged from 0·6° to 2·8°. This is the only harmonic showing a clear difference between high speed and low speed trials. The high speed amplitudes are greater and the low speed ones are lower than 1°;

(c) some of the tests indicated a significant contribution of the second harmonic which was largely scattered in phase. The corresponding sector was not drawn because the most part of the tests did not show these harmonics.

II) Amputees

Both the frontal and horizontal pelvic rotations had relevant extrinsic harmonic contributions. Concerning the rotations in the *frontal plane;* (a) the first harmonic covered a sector of the polar plane centred at 150°, the amplitudes ranged from 2° to 5·4°;

(b) the third harmonic covered a sector shifted in phase with respect to the normal one, while the amplitude range was similar for both categories of subjects;

(c) the second (extrinsic) harmonic was always significant and had an amplitude from 0·5° to 1·5° and a certain amount of within-subject repeatability.

With reference to the *horizontal plane;* (a) the first harmonic in the amputee tests fell in a sector ranging in phase from −30° to +50°, with amplitudes varying from 2·2° to 7·0°.

(b) the third harmonic covered a sector centred on −50°, with amplitudes from 0·7° to 1·7°, and comparison with normal subjects showed a neat phase opposition;

(c) the second harmonic had amplitudes between 1·1° and 3·5° and showed within-subject repeatability;

(d) in few cases only the fourth harmonic contribution was significant. Thus, it was not reported in the figure.

Shoulder rotations (Fig. 5)

I) Normal subjects

In normal subjects, both the horizontal and the frontal rotations are in accord with descriptions referred to in the literature (Murray et al, 1964). The Fourier analysis showed that these rotations are practically sinusoidal. All harmonics other than the first can be disregarded, since their amplitude was less than 0·5°. In the *frontal plane* the sector covered by the first harmonic ranged in phase from −80° to −185° and in amplitude from 0·5° to 3·5°. Comparison with the same rotation of the pelvis in Figure 3, showed that the shoulders and pelvis of normal subjects rotate in counterphase. On the horizontal plane the sector was centred on −70° of phase and the amplitudes ranged from 2° to 8°. Owing to the phase scatter of the corresponding pelvic harmonics, no phase relationship can be recognized.

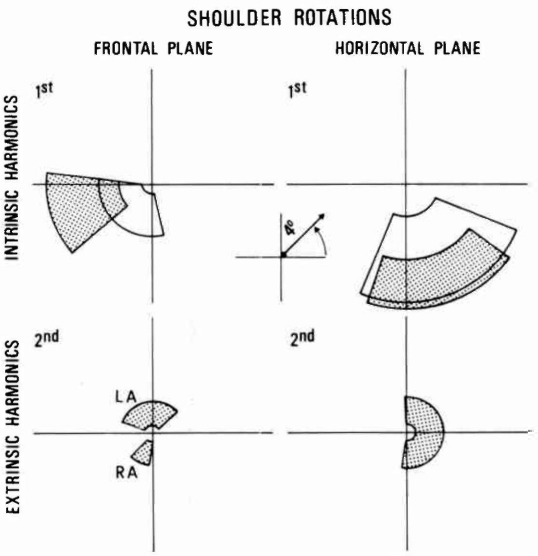

Fig. 5. Synthetic representation of the shoulder horizontal and frontal rotations in normal subjects and amputees: intrinsic and extrinsic harmonics are separately drawn. See also Fig. 3. LA = left amputation; RA = right amputation.

II) Amputees

The extrinsic (second) harmonics cannot be disregarded in either rotation. In contrast, the third harmonic can be disregarded. Moreover, in the *frontal plane;* (a) the first harmonic covered a sector more limited in phase than in normal subjects, with a phase range from −140° to −190° and amplitude variations from 1·9° to 7°;
(b) the second harmonic had phase values which were dependent upon the amputation side.

In the *horizontal plane;* (a) the first harmonic amplitude was larger than in normal subjects, phase values fell between −35° and −110° and amplitude from 5° to 8·5°;
(b) the second harmonic occupied the first and fourth quadrant of the polar plane with amplitudes ranging from 0·6° to 2·5°.

Trunk rotations (Fig. 6)

I) Normal subjects

In the *frontal plane* the rotation was described by the first harmonic only. Its phase varied from 30° to 110° with amplitudes from 0·6° to 2·5°.

In the *sagittal plane* rotation the intrinsic (second) harmonic covered a narrow sector centred on 160° of phase with amplitudes from 0·5° to 2°. This rotation had a first (extrinsic) harmonic that was significant, but with a large scatter on the phase, that covered 2 radians. Neither intraindividual nor speed dependent trends were recognized.

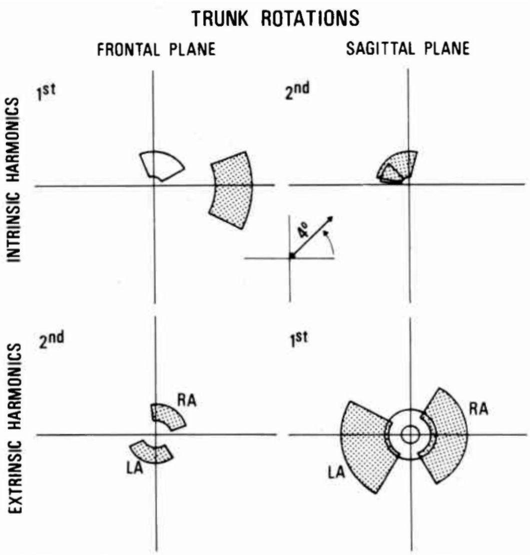

Fig. 6. Synthetic representation of the trunk sagittal and frontal rotations in normal subjects and amputees; intrinsic and extrinsic harmonics are separately drawn. See also Fig. 3. LA = left amputation; RA = right amputation.

II) Amputees

In both planes the extrinsic harmonics are larger than in normal subjects; (a) the first harmonic of the *frontal rotation* had amplitudes that ranged from 4° to 6·8° and occupied a sector with phase values between −30° and +20°;
(b) the second harmonic of the same rotation depended upon the side of amputation and its amplitude varied between 1° and 2°.

Concerning rotation in the sagittal plane, the extrinsic (first) harmonic became dominant; its amplitude went from 1·2° to 5° and the phase depended upon the side of amputation. The intrinsic (second) harmonic covered a larger sector than that of the normal subjects; this sector was centred on − 130° of phase values. The amputee's second harmonic had an amplitude ranging from 0·5° to 2·4°.

Discussion

Since the normal tests considered in this work did not show significant variations with the progression speed, it seemed appropriate to collect all of them in one control sample for the comparison with amputee's data.

Amputee gait, compared with normal gait, is characterized by remarkably larger amplitudes of the harmonics in all rotations we investigated. Furthermore, extrinsic harmonics appear. This means that the amputee's upper body rotations are larger and asymmetrical.

Concerning the first harmonic of the pelvic frontal rotation in normal subjects, it corresponds to the fall of the pelvis on the side of the swinging leg, referred to by Saunders et al. (1953) as a gait determinant. In the amputee tests this harmonic is in counterphase with respect to normal, which means that there is an elevation of the pelvis, as opposed to a fall, on the side of the swinging leg. Such amputee behaviour can be correlated with the passive prosthetic ankle and the reduced efficiency of the stump abductor muscles on the amputated side. During prosthetic swing the artificial foot cannot be dorsiflexed; thus the corresponding hip must be elevated in order to gain clearance for the swinging leg. During prosthetic stance the sound hip is elevated and the trunk bends laterally toward the prosthesis in order to make equilibrium easier and to decrease the effort of the stump abductors (McLeish and Charnley, 1970).

The overall trend of the pelvic horizontal rotation may be described by the first harmonic alone. According to the relevant description given by Saunders et al. (1953) or by Steindler (1955), its phase values should fall between +90° and + 180°. Actually the normal subjects' tests in this study showed the above feature only exceptionally and exhibited a very large inter-individual phase scatter (Fig. 4). This suggests that the pelvic horizontal rotation is an individual trait.

As far as the amputee is concerned the first harmonic of the pelvic horizontal rotation belongs to a sector centred on a phase value near zero. This means that the amputee's larger forward rotation of the pelvis occurs during mid-stance. This behaviour can be correlated with the structural and functional losses of the amputee. The forward movement of the normal hip during prosthetic stance is opposed by knee stability problems, reduction of the stump muscles efficacy and prosthetic ankle passivity. Because of the absence of an active push-off by the prosthesis, the pelvic rotation is reduced during normal leg stance.

With regard to the third harmonic of the pelvic horizontal rotation, the phenomena associated with it are not easily detectable. A third harmonic component may be engendered by two perturbations, equal and opposite, superimposed on a curve with period T, if the interval between the perturbations is T/2. The amplitude and phase of the third harmonic depend on the amplitude, duration and location within T of the perturbations. By inspections of the horizontal pelvic rotation plots vs time such perturbations can be only indistinctly seen, because of the small amplitude of the third harmonic itself and the large phase scatter of the first harmonic. If X be the mean direction of progression of the subject with respect to the laboratory frame, in a reference system moving with the subject itself at a medium speed of progression, then the plot of the difference between the X coordinates ($\Delta X(t)$) of the hip, purged from the contribution of the first and second harmonics, shows the perturbations very distinctly*. In Figure 7(a) a typical plot for a

*It can be easily proved that due to the characteristics of the hip movements, the pelvis geometry and the formal definition of the pelvic horizontal rotation, the perturbations of the angle are markedly reduced with respect to the corresponding $\Delta X(t)$ perturbations.

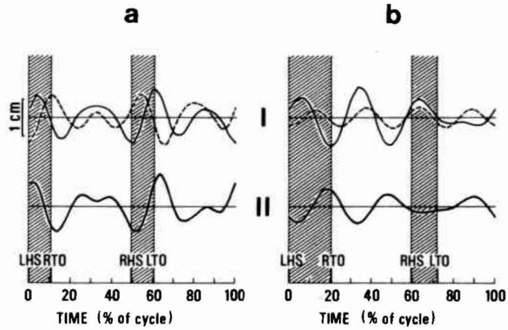

Fig. 7. $\Delta X(t)$ plots relative to a normal subject (a.II) and an amputee (b.II). In (a.I) the X(t) plots purged by the first and second harmonics' contributions, of right (continuous line) and left (dotted line) hip are shown. In (a.II) the large perturbations located in correspondence with the double support phases are evident. Larger contributions from the leg entering its swing phase are noticeable. In (b.I) the same quantities than in (a.I) relative to the amputated leg hip (continuous line) and to the sound leg hip (dotted line) are represented. In (b.II) the perturbation during double support is confined to the sound leg restraint and is opposite to the corresponding normal one. The larger values of perturbation happen during sound leg support and depend on the movement of the prosthetic hip.

normal subject is shown together with the corresponding hip coordinates. It is clear that the perturbations are located in correspondence with the double support phase, and that the larger contribution of each of them is due to the limb entering its swing phase. Both these features are common to all tested subjects. It can be verified that such perturbations engender a third harmonic with a phase value corresponding to that actually exhibited by normal subjects. Consequently it appears that the third harmonic corresponds to rapid pelvic movements depending upon the mechanisms operated during the double support, in particular during push-off for each leg. Based on the repetitive quality of the characteristic described by the third harmonic of the horizontal rotation of the pelvis, one can conclude that it is an invariant of human locomotion. As shown in the previous section under Results, this characteristic changes in the amputee's ambulation; in fact, the third harmonic was negligible in three of the tests, and had negative phase values in the others. For the sake of brevity only the negative phase results will be discussed. In Figure 7(b) an example of the corresponding $\Delta X(t)$ plot is shown. A perturbation appeared opposite to that of the normal subject during double support and had its maximum value during sound leg support. This behaviour seemed to be related to the following handicaps of the amputee; passive prosthetic ankle, which implies that the prosthesis is unable to push the hip forward during deploy; and prosthetic knee stability, for the sake of which the backward movement of the prosthetic hip during early stance is inhibited. A pelvic contribution to the prosthetic swing can be hypothesized, probably as a compensation for the passive push-off of the prosthesis.

With reference to the third harmonic of the pelvic frontal rotation, a discussion, parallel to that of the third harmonic of the pelvis horizontal rotation, could also be carried out. It should be noted that in normal subjects the third harmonic of the pelvic frontal rotation may be related to a perturbation occurring during the double support; thus it can be viewed as corresponding to an invariant of locomotion.

Conclusions

Two kinds of conclusions can be drawn from the above discussed results; (a) the usefulness of an analysis of the upper body segment rotations by means of harmonic components for the purpose of gait evaluation; (b) to improve prosthetic design some hypotheses concerning the mechanisms may be related to specific characteristics of the amputee's movement. The initial assumption that the upper body segment rotations can facilitate an assessment of the differences between the normal and the amputee's gait has been confirmed. In particular, substantial quantitative differences appear at pelvis level. Through the use of harmonic components more clear-cut identification and comprehension of these differences was established.

In that the above differences are apt to be correlated with increased metabolic costs of amputee gait and with increased mechanical loads on the spine, a focus on the upper body movements for the purposes of gait evaluation is justified. The representation of the harmonic components in polar plots seems to be particularly valuable due to its ability in discriminating between normal and abnormal characteristics.

With regard to the compensatory mechanisms the amputee has to perform and their tentative relationships with specific deficiencies of present day prostheses, the above analysis suggests a need for some active mechanisms in the prosthetic knee and ankle. These mechanisms should be devoted to ensuring a hip movement as similar as possible to the normal one, assuming that it defines an optimum condition (Cappozzo et al. 1979b). Possible suggestions for the design of such mechanisms are; (a) a knee-ankle mechanism able to dorsiflex the foot during knee flexion (Cappozzo et al. 1980), thus reducing hip elevation during prosthesis swing; (b) a knee equipped with some kind of energy recovery mechanism able to perform suitable knee flexion-extension during the early stance phase in order to reduce knee stability problems (Cappozzo et al, 1979b; Seliktar, 1971). Such mechanisms should allow an improvement of horizontal pelvic rotation.

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