KINEMATICS OF SPRINTING: COMPARISON BETWEEN NORMAL AND AMPUTEES ATHLETES

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The paper introduces the initial activities of a project related to the kinetic analysis of amputees athletes running by means of dedicated orthopaedic prostheses. In this first session it has been analysed the kinematics of four athletes sample, three normal (the first a high-level pro, the second a junior and the third an amateur) and one unilateral below-knee paralympic amputee. Research objective is to analyse the kinematics of the amputee athlete and the differences in respect of normal athletes. It has to be considered that each normal athlete has its own way of running which depends on his anthropometrical parameters and on its motion strategy: an ideal biomechanics of running, which is a reference to optimise the performance of athlete, doesn't exist. First of all it has been described the running biomechanical phases of normal athletes and the structural components of a sprinting prosthesis; then the system to analyse human movement and finally the adopted protocol for acquisition. The kinematical parameters and the motion strategy adopted by amputee are different in comparison to normal athletes: it has been analysed the anthropometrical parameters, the time-space parameters and pelvis, hip, knee and ankle angles. The collected data will be applied to the design process of running prosthesis in the socket manufacturing and in the alignment of prosthetic components to optimise the performance of the athlete.

KEY WORDS: motion capture, amputee running, prosthesis, sport performance.

INTRODUCTION: Sprinting is a type of running, performed over shorter distances and at faster speeds. The goal of sprinting is to cover the short distance (100m –200m) in the shortest time. The gait cycle is the basic unit of measurement in gait analysis and begins when a foot comes in contact with the ground and finishes when the same foot returns to the ground. It includes two principal phases: the stance phase and the swing phase. The duration of stance phase, as a percentage of stride time, decreases, proportionally with the speed, from around 60% during the walking, to 30% for running and to 22% for sprinting. In sprinting, with high speed, initial contact changes from hind foot to forefoot. During initial contact the sprinter contacts the ground and the Centre of Gravity takes the lower position, the hip is flexed, the knee is extended and the ankle is in dorsi-flexion position. From initial contact (IC) to mid-stance (MD) the sprinter absorbs energy ("absorption phase"). The propulsive forces are anterior to the joints of hip and knee, while it is posterior to ankle where it generates a plantar flexion moment. From mid-stance to toe-off (TO) starts the "generation phase", where the body generates the propulsive forces for advancement in sprinting and at toe off moment the foot finishes the contact with the ground. The generation phase ends at mid swing frame (MS). From mid-swing the lower limb begins to decelerate till terminal swing (TS). At the begin and at the end of swing phase has to period of double-float, where neither limb is in contact with the ground: from walking and running to sprinting the stance phase decreases and increases the float phase. The amputee athlete uses a dedicated sprint prosthesis with Springlite Foot by Otto Bock (Fig.1). The prosthesis is made of three principal components: the liner, the socket and the foot. The liner, the interface between socket and stump, has to protect the stump from injuries and loads that the stump suffers during running. It is made of soft and elastic material and the level of stump comfort depends on this. The socket is the custom made prosthetic component, obtained through the plaster cast on the stump. It has to allow the complete knee rotation in order to optimize the running, but at the same time it has to guarantee the secure anchorage to the knee.(Fig.2) The foot's material is carbon fiber, produced through technologies from aeronautic and military industry The foot represents the active prosthetic component: it stores and releases the energy and it reduces the stump traumas. The running foot efficiency has around 80 percents. In comparison, the human foot has 241 percents spring efficiency, with the addition of the concentric plantar-flexion contraction. The running foot "J" shape design allows to controll dorsi-flexion and knee flexion: the prosthetic foot stores energy as the sound foot through the Achilles tendon and the arch of the foot. The CF foot design has allowed the amputee athletes to reach a high competitive level: the 100m world record below-knee amputee is 11,07s.

Fig.1 Springlite Foot

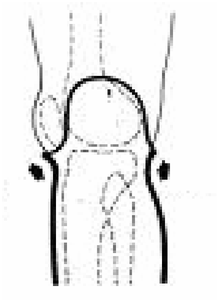

Fig.2 Socket

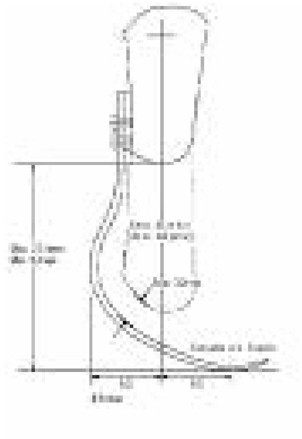

Fig.3 Alignment Socket-Foot

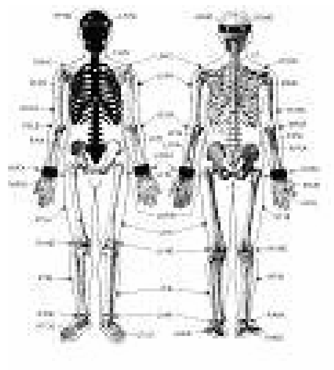

Fig.4 Helen Hayes marker set

METHOD:

Data Collection: the optoelectronic VICON system measured the three-dimensional coordinates of the markers, elements of reflecting material that are sticked on the body of the subject on anatomical landmarks. In sprinting the sampling frequency must be higher, but not so much to avoid creating noise. While in gait analysis six cameras of 50-100 Hz frequency around 5 m long are used, in sprinting 8 infrared cameras of 100-400 Hz are used. The antagonist positioning of the cameras creates a virtual volume inside which the software captures the signals reflected by the markers positioned on landmarks of the athlete (each camera has to capture at least two markers and each marker has to be visible from at least a couple of cameras).The performance area is 12 long and 4 meters deep. The maximum height covered is 2 meters and it is chosen according to the height of the athletes. Once known the three-dimensional coordinates of the markers it is possible therefore to calculate trajectories, kinematical angles and the sprint speed. Initially static calibration of cameras begins through shooting a point grid set orthogonally in the middle of performance area. In a second moment takes place space dynamic calibration inside the performance area by waving a marker wand quickly. The next step is the measurement of athletes anthropometrical parameters according to protocol Kadaba-Davis (Table 1). This method allows to suit the standard virtual model to the examined athlete. The subject is asked to maintain the orthostatic position for around 5 sec in the centre of the performance area in order to verify the correct survey of all the markers through all the cameras. The acquisition consists of the following tests: three skips; three jumped runs; three launched runs or sprint. We have chosen these series of session to verify the posture and the motion strategy of athletes in extremes condition. The markers, positioned according to the Newington/Helen Hayes marker set, are attached on the following landmarks (Fig.4): head, shoulders, elbow, upper arm, wrist, 7^{th} cervical vertebrae, 10^{th} thoracic vertebrae, clavicle, sternum, right back for the upper parts of human body. LASI, RASI,LPSI and RPSI in the pelvic, placed directly over the superior iliac spine; LKNE, placed on the lateral epicondyle of the left Knee; LTHI, left Thigh, placed over the lower lateral 1/3 surface of the thigh, although the high is not critical; LANK, left ankle, placed on the lateral; LTIB, left tibial wand marker, similar to the thigh markers, these are placed over the lower 1/3 of the shank to determine the alignment of the ankle flexion axis; LTOE, placed over the second metatarsal head; LHEE, left heel, placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker; the same markers for right arm, hip and leg. The two markers, positioned sideways on the superior iliac spine, are redundant, so that, if during the running, the some markers, could be lost by the athlete, it is possible to reconstruct the model .For landmarks on the prosthesis it has been followed Buckley-method with the amputee on tip-toe: epicondyle on the socket, malleolus, heel, tip of feet and second metatarsal virtual. To analyse the foot behaviour and to verify the right alignment between socket and foot (Fig.3), it has been attached the markers on the following points: one on the under patella zone, two vertically on T connection, two on the foot at the same height on T.

Data Analysis: It has been examined the time-space parameters (Tab.2) and rotation joint angles of lower limb for four athletes sample, three normal and one unilateral below-knee paralympic amputee (the grey band describes the run of normal athlete and the thin centreline shows the average, the red and green line describes the run of left and right limb of only one amputee) : ankle, knee, hip and pelvis (Fig.5). Each graph shows the sprint cycle of both limbs starting from the initial contact to contact to the ground with the same foot. The condition in which the segments thigh and shank are aligned and perpendicular to the ground, during the stance phase, it is considered like zero reference. In this position, when the ankle rotates up, the positive angle refers to ankle dorsi-flexion, otherwise is plantar flexion. The knee angle in complete extension equals zero and each change is considered shows the knee flexion. The angle described by the thigh, in respect to the perpendicular to pelvis plane, is positive and it shows the hip flexion during the advancement in sprint. On the frontal plane it is considered positive the hip adduction when the thigh approaches the perpendicular to pelvis plane, otherwise is considered hip abduction. As far as the pelvis is concerned positive the obliquity when the pelvis rotates up on the frontal plane, the pelvic tilt when the pelvis is lowered down on the sagittal plane and the rotation when the pelvis intrarotates.

RESULTS:

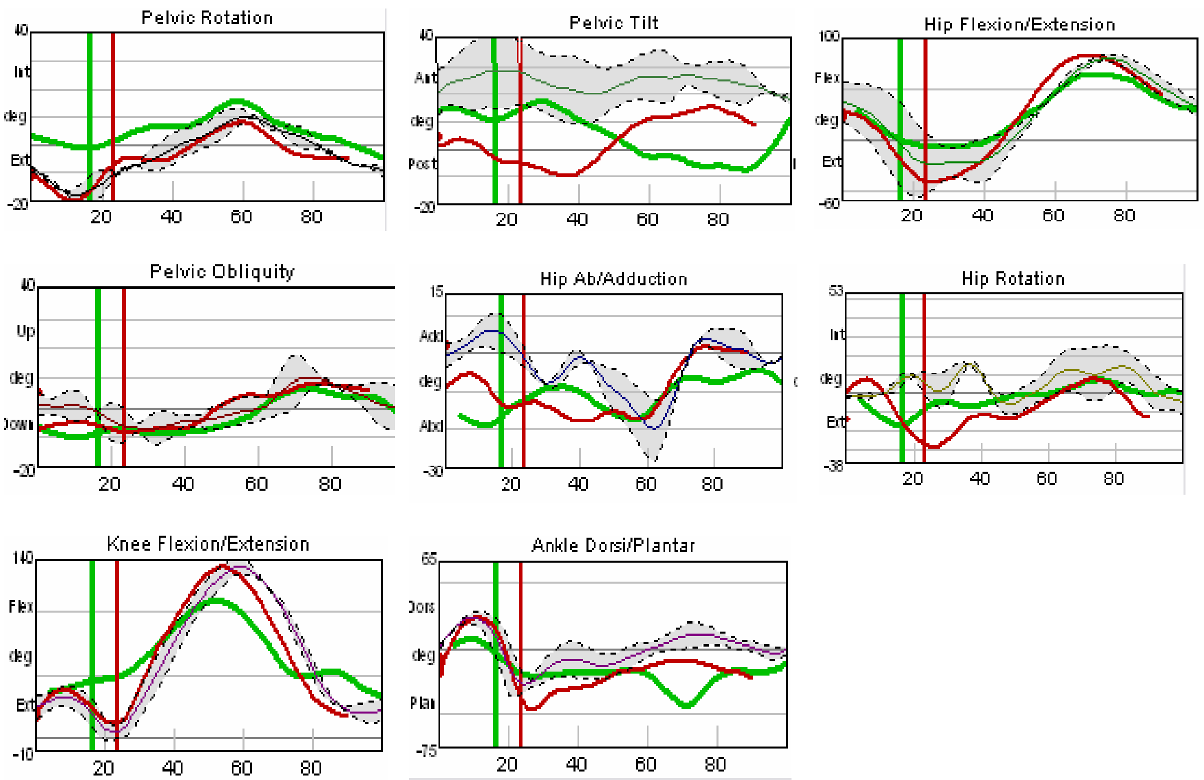

Fig. 5 Angles pelvis, Hip, knee, ankle

Athlete	Pro		Junior		Amateur		Amputee	
Birth Year	83		87		90		71	
Personal Best (s)	10,45/21,13		16,95(110hs)		38,3(300m)		12,44/ 25,12	
Height (m)	1,83		1,85		1,74		1,80	
Weight (Kg)	78		83		68		72	
Lower Limb antropometrical Parameters	R	L	R	L	R	L	R	L
diam.Knee (10^{-2}m)	10,0	9,9	10,5	10,0	11,0	10,0	10,0	10,0
diam. Ankle (10^{-2}m)	7,5	7,5	8,0	8,0	7,2	7,2		7,5
Leg (10^{-2}m)	98	98	99	99	93	93		93

Tab.1 antropometrical parameters

Time and Space Parameters	Pro		Junior		Amateur		Amputee	
	R	L	R	L	R	L	R	L
Width Step (m)	0,21	0,21	0,15	0,12	0,06	0,08	0,36	0,32
Lenght Stride (m)	4,58	4,61	4,16	4,2	3,78	3,88	3,92	3,92
Stance Phase	20%	20%	21%	21%	22%	24%	22%	18%
Swing Phase	80%	80%	79%	79%	78%	76%	78%	82%
Cadence (steps/min)	264		240		250		238	
Sprint Velocity (m/s)	10,2	10,0	8,32	8,41	8,05	7,50	7,85	7,68

Tab. 2 time-space parameters

DISCUSSION: The asymmetry, between the prosthetic and the sound limb, is really remarkable: in fact, while the swing phase of the sound limb lasts the expected 22% of sprint cycle and so in line with the percentage of the referring normal athletes, the prosthetic limb is just a little lower than 20%. It has been underlined the compensatory mechanisms of sound limb due to the different kinematics of prosthetic limb. The ankle rotation of the sound limb plays as a rule . A weak dorsiflexion appears at the beginning of the stance phase for the prosthetic ankle due to the foot elasticity and shape; the nearly horizontal feature during the swing phase is due to foot stiffness and its morphology creates a remarkable plantar flexion. The sound limb knee evidences a sinusoidal trajectory as normal athletes, with the advance close to the maximum flexion peak (140°) due to a compensatory mechanism of the prosthetic limb: the socket does not allow a knee flexion more than 110°. The prosthetic limb knee is not able to complete knee extension before ground contact since the alignment between socket and foot always maintain a flexion knee approximately around 30°. The hip rotation graph of the sound limb on the three planes plays as a normal athletes with the exception of an advance in the maximum flexion peak on the sagittal plane due to a compensatory strategy as in the knee case. The prosthetic limb hip evidences an extension next to the zero, smaller than the extension hip of the normal athletes (30°) and a smaller flexion. On the frontal plane, during the stance phase, the prosthetic limb hip goes in adduction, contrarily to the sound limb, in order to compensate the knee intra-rotation. The pelvis angles are calculated in the absolute reference, while all the others are relative angles. In pelvic obliquity graph the pelvis is lowered on the prosthetic limb side and its value becomes the minimal peak at the moment of maximum loading on the prosthetic foot which acts as a spring. During the stance phase, on the horizontal plane, the hip intrarotation compensates the hip abduction in order to allow the advancement in sprinting. The Range of Motion of pelvic tilt is proportional to the athlete energetic consumption since sprinter trunk has to be constantly ahead: the trunk of normal athletes is more advanced, therefore the RoM of pelvic tilt is narrower. In the swing phase the pelvis, referring the prosthetic limb, shows an trunk regression due to hip and knee weak flexion.

CONCLUSION: Next step is to enrich the sample with other amputee athletes. Sprinting is the result of multiple factors: the prosthesis influence, determined by alignment of prosthetic components and by foot through elasticity and shape; the motion strategy and the compensatory mechanism of the sound limb; the period of session acquisition during winter training program with an unoptimal athletic condition of subjects and an environmental condition. It should be interesting to repeat acquisition sessions along different training periods in order to evaluate precisely the impact of the athletic condition. The results by the analysis will be used to manufacture a socket which has to allow a better knee flexion and a optimisation of socket-foot components alignment in order to exploit the maximum foot energy.

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