A SIMPLIFIED CAD MODEL OF TRANS-RADIAL SOCKETS
SUITABLE FOR SUBJECT ENERGY EXPENDITURE ASSESSMENT

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1. INTRODUCTION
The measurement of the mechanical energy expenditure of a subject in accomplishing a given motor task has been reported as a valuable index to quantitatively assess his/her motor ability or pathology stage [1-3]. In order to compute this parameter applying inverse dynamics techniques [1], the subject’s joint kinematics during the motor task and the inertial parameters (i.e. the mass, centre of mass and inertia matrix) of his/her moving limbs must be known. While the joint kinematics can be measured in-vivo using a motion analysis system (e.g. an optoelectronic system), the inertial parameters of human limbs are usually retrieved from anthropometric tables [1]. When the subject acquired is an amputee fitted with an artificial limb the problem of determining the prosthesis inertial parameters therefore arises. Considering a trans-radial amputation levels, a prosthesis is composed by standardized parts (hand, battery, lamination ring and actuators), and the subject-specific inner and outer sockets: since the CAD models of these parts are usually unavailable their single and cumulative inertia parameters remain unknown. The aim of this work was therefore to propose two possible simplified CAD models for trans-radial prostheses and to identify among them which one leads to the best estimation of the mechanical energy expenditure during a flexion-extension of the elbow in the sagittal and horizontal plane and during a shoulder internal-external rotation. The models, of increasing complexity, take into account the sockets, battery and lamination ring.

2. MATERIALS AND METHODS
2.1 Gold standard
As a gold standard and basis to develop the two simplified CAD models, the inner and outer sockets of three amputees where digitalized using a high-accuracy laser scanner (Minolta VIVID 900, resolution: 25pts/mm², accuracy: 0.25mm). Since the scanner was unable to fully digitize the inside of the sockets, only the external surfaces were retained from the scans. For all the three prostheses, measurements on the assessable parts showed a 3mm thickness for the inner and 2mm for the outer sockets. These offsets were then applied to the external surfaces to obtain the solids shown in Figure 1. Using the CAD software Rhino 3.0 (Robert McNeel & Associates), the inner parts were visually fitted into the corresponding outer parts, inspecting parallel sections of the solids to exclude interpenetrations. The obtained inner-outer sockets

![Gold standard models GS1 (left), GS2 (middle) and GS3 (right), this last shown with lamination ring and battery.](image-url)
couples will be referred hereinafter as DS1, DS2 and DS3. For each couple, a hollow cylinder (d_i= 3.6cm, d_e= 5.4cm, h=2cm) was generated distally to model the lamination ring Otto Bock 10S1. For models DS1 and DS3, a parallelepiped of 1.5cm height was generated into the battery housing digitized in the scans to approximate the battery Otto Bock 757B21; this was done by “virtual palpating” on the CAD models the three corners of the battery housing to define the base of the parallelepiped and then imposing the 1.5cm height. No battery model was added to DS2 since for this type of sockets no fixed battery position is prearranged. The sets formed by a DS, a lamination ring and a battery (where applicable) will be referred hereinafter as GS1, GS2 and GS3.

2.2 Simplified CAD models design steps

Given the high complexity of the shapes of the sockets, we developed two simplified models characterized by increasing details: a first order approximation based on truncated cones (TC) and a second order based on the combination of truncated cones and tubular shapes (TS).

The conceptual design steps were the same for both: 1) definition of a set of easily identifiable geometric landmarks on the outer socket only; 2) “virtual palpation” of these landmarks on GS1-GS3 in Rhino; 3) creation, for every GS, of TC/TS design curves through these points; 4) application of solid modelling techniques to obtain TC/TS, generating both an outer and an inner socket model; 5) generation of the lamination ring into the models; 6) generation of the battery. Since the two models are were generated directly on every GS, there was no need to actually perform steps 4) and 5) since the lamination ring and battery models would have been the same generated for the corresponding GS. In the following two sections the basic features of TC and TS are given. Further details can be found in [4].

2.3 Generation of the TC model

The 7 landmarks selected for the construction of TC are shown in Figure 2a: A-B-C on the wrist edge forming a circle, D at the most distal point of the slot between the outer and inner socket, E-F where the lateral and medial elbow epicondyles are located, G at the point of maximal curvature on the proximal left or right crest of the socket, depending on which of them is the furthest from the wrist. The outer socket was obtained by designing an hollow truncated cone (2mm thick) using points A-B-C-E-F-G (Figure 2b). The inner socket was derived from the outer by offsets (3mm think), adjusting its length to have D as the most distal point. Only the distal bottom of the inner socket was closed. Taking into account the construction of the battery, TC required the identification of 10 landmarks. The construction of TC was realized for all the GSs, obtaining TC1-TC2-TC3.

Figure 2 Selected landmarks for TC construction (a) and TC outer socket for GS1 (b)
2.4 Generation of the TS model

The 15 landmarks selected for the construction of TS are shown in Figure 2a and 3a: A-B-C-D are the same used for TC; E-F-G form a circle approximately parallel to the A-B-C cycle but at the edge of the stump insertion hope; M-N-O-P are locate at the top of the corners of the proximal aspect of the socket; H-I lie at the end of the straight edge forming the slot for the olecranon; L lies midway on the socket edge between E and M. The A-B-C and E-F-G cycles were used to build the hollow truncated cone of the outer socket. The E-F-G, E-H-I and L-M-N cycles were used to build the tubular shapes forming the proximal part of the outer socket. The same points were used to define 3 cutting planes to design the olecranon slot. After imposing a 2mm thickness, the solid of the outer socket shown in Figure 3b was obtained. The inner socket was derived from the outer, imposing a 3mm thickness, and adjusting its length to have D as the most distal point. Only the distal bottom of the inner socket was closed. Taking into account the construction of the battery, TS required the identification of 18 landmarks. The construction of TS was realized for all the GSs, obtaining TS1-TS2-TS3.

2.5 Outcome measures

For the comparison, each triplet (GS-TC-TS), i=1,2,3, was set in the Rhino global reference frame (G) to resemble a forearm when the elbow is flexed 90°, with the palpated lateral and medial epicondyles laying on the G_y axis, and the normal to the A-B-C circle parallel to the G_{xy} plane (Figure 2b). In this configuration, G_y replicated the elbow flexion-extension axis [5]. For each triplet, the mass, centre of mass and inertia matrix with respect to G of GS_i, TS_i and TC_i were computed using the Rhino “Mass property” function and custom-made MATLAB (The Mathworks) functions. A mean density of 1109 Kg/m^3, 900 Kg/m^3 and 1897 Kg/m^3 was assumed (based on measured weights and volumes) for the sockets, lamination ring and battery, respectively. Using these data and applying the torques law of dynamics, the energy mechanical expenditure (MEE) [1] during an elbow complete flexion-extension in the sagittal (FLEX) and horizontal (FLEXh) plane was computed and that of TS_i and TC_i was compared to the one of GS_i. MEE was obtained by integrating the absolute value of the power consumption over the simulation time. In particular, FLEX and FLEXh were simulated by imposing a cosinusoidal rotation (amplitude: from -70° to 70°, frequency: 280°/s) for 1 second, around G_y considering, for FLEX, and not considering, for FLEXh, the gravitation force along G_z. Finally, the MEE was also computed for a shoulder internal-external rotation movement (INEX), assuming the elbow fixed at 90°. This movement was simulated by applying the same motion described above around G_z.

3. RESULTS AND DISCUSSION

Results for the different MEEs are reported in Table 1. As can be observed, TS errors in MEE computation ranged from 2% to 8.5%, with best performances for FLEXh and INEX, i.e. in the trials where the inertial forces, and thus the estimation of the inertia matrix, were the most important factors. On the contrary, TC gave the best results where the gravitational force was considered, due to its systematic underestimation of the gravity lever arm (-0.5%– -2%) which
compensated the systematic overestimation of the mass (2.5%–12%). TS always overestimated the lever arm (0.7%–1.3%) and the mass (5%, 7.5%), with the exception of the second triplet (mass=-2.5% and lever arm: +7.8%). The MEE errors of TC ranged, on the whole, from 2.5% to 12%. TS thus ensured the overall best performances. However, the use of TC appears to be the best practical choice, considering 1) the time required to digitize the 18 landmarks of TS, 2) that the differences from TC and TS are limited to 4% in the worst case, 3) the absence from the model of a prosthetic hand, which - being about 500g located distally - will reasonably reduce the percentage MEE error.

The generation of every TS, and TC, directly on the respective GS excluded the problem of the spatial registration of the models on the gold standard. It is important here to notice that the procedure used for the models generation is applicable even if no high-accuracy models are available. The “palpation” of the landmarks can in fact be easily done using a stereophotogrammetric system directly on the real prosthesis, either applying micro-markers on the landmarks, or using the CAST technique [6]. The procedure thus appears to be applicable in routinely motion analysis.

<table>
<thead>
<tr>
<th>GS1 [J]</th>
<th>TC1 % error</th>
<th>TS1 % error</th>
<th>GS2 [J]</th>
<th>TC2 % error</th>
<th>TS2 % error</th>
<th>GS3 [J]</th>
<th>TC3 % error</th>
<th>TS3 % error</th>
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<td>2.51</td>
<td>6.01</td>
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<td>10.23</td>
<td>4.79</td>
<td>1.55</td>
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<tr>
<td>FLEXh</td>
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<td>4.07</td>
<td>5.39</td>
<td>0.64</td>
<td>9.18</td>
<td>3.94</td>
<td>0.75</td>
<td>9.03</td>
</tr>
<tr>
<td>INEX</td>
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<td>4.64</td>
<td>0.68</td>
<td>10.94</td>
<td>2.09</td>
<td>0.83</td>
<td>11.89</td>
</tr>
</tbody>
</table>

Table 1 Results for the MEE computation for the three prostheses. Results for TCs and TSs are expressed as percentage errors of the values reported for the corresponding GSs.

4. CONCLUSIONS

The aim of this work was to identify the level of detail required in modelling a trans-radial prosthesis (formed by the socket, the lamination ring and the battery) to properly compute the mechanical energy expenditure of a subject during a flexion-extension of the elbow in the sagittal and horizontal plane and during a shoulder internal-external rotation. The two simplified model proposed, when compared to gold standard models, proved to reproduce the real mechanical energy expenditure within a 12% error for the less detailed model and 8% for the most detailed, in the worst case. Given these limited difference, the simpler model appears to be most practical choice for energy expenditure assessments. Future developments will be intended to evaluate the effect of a prosthetic hand on the models, and to measure the differences in the estimated mechanical energy expenditure at the shoulder level during activities of the daily living.

5. ACKNOWLEDGMENT

The authors are grateful to ENEA and Ing. Sergio Petronilli for having managed the laser scanner acquisitions.

6. REFERENCES