THE MYOKINETIC CONTROL INTERFACE: HOW MANY MAGNETS CAN BE IMPLANTED IN AN AMPUTATED FOREARM? EVIDENCES FROM A SIMULATED ENVIRONMENT

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ABSTRACT

The displacement of residual muscles during voluntary contraction in a transradial amputee could be effectively exploited to control multiple degrees of freedom in a hand prosthesis. We recently introduced a new human-machine interface (the *myokinetic control interface*) which aims at tracking muscles contraction through implanted permanent magnets and magnetic field sensors located inside the socket. Magnetic markers (MM) tracking systems have been widely investigated in the past, especially for controlling and guiding medical tools for intra-body applications. However, specific design rules for a multiarticulate robotic hand control system have not been defined yet. Here, we studied the tracking accuracy of multiple implanted magnets by simulating different levels of trans-radial amputation using a 3D CAD model of the forearm. A magnets placing procedure was developed to position the MMs in the available muscles, following general guidelines derived in our previous study. The localizer was able to accurately track up to 9, 13 and 18 MMs, in a proximal, middle and distal representative amputation, respectively. Localization errors below ~3% the length of the trajectories travelled by the MMs during muscles contraction were achieved for all amputation levels. Not only this work answers the question: "how many magnets could be implanted in a forearm and successfully tracked with a myokinetic control approach?", but also provides interesting insights for a wide range of bioengineering applications exploiting remote tracking.

INTRODUCTION

An upper extremity amputation is an event that profoundly affects the quality of life in several aspects, limiting the individual in performing working and daily living activities. Commercially available artificial hands and arms are often controlled through surface EMG electrodes that record the electrical activity generated by the residual muscles when contracting. Often, this approach suffers the lack of accessible independent control sources, and its performance is thus limited in the case of multi-articulated prostheses [1]. In the last years, different solutions have been proposed to overcome this limitation and increase the number of degrees of freedom (DoFs) that can be controlled independently.

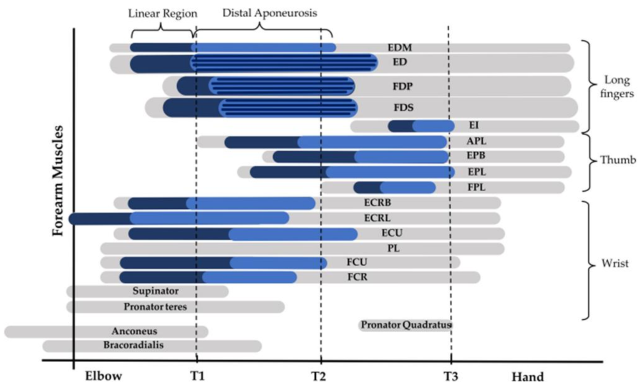

Figure 1: Detail on the anatomical distribution of the extrinsic muscles of the hand inside the forearm. The colored sections indicate the regions of the muscles belly which were considered for the implant. Dark blue regions indicate the proximal sections, which contract proportionally to their distance from the muscle origin. Light blue regions indicate the distal sections, which move by an amount corresponding to the maximum physical contraction. Gray muscles/sections were instead excluded. The three simulated levels of amputation (T1, T2, T3) are also indicated.

New technologies like wireless implantable myoelectric sensors (IMES) [2] or epimysial electrodes wired through osseointegrated implants [3], enabled direct interfacing with the physiological structures involved in the motor control, when still intact. Our group recently proposed a new concept of human-machine interface for the control of artificial limbs that takes advantage of magnetic tracking, termed *myokinetic control interface* [4]. The idea is to implant multiple permanent magnets (magnetic markers – MMs) into the residual muscles of an amputee, track their movements using magnetic sensors hosted in the socket, and use these signals as control inputs in a prosthesis, e.g. a hand. Notably, localizing the implanted magnets is equivalent to measure the contraction/elongation of the muscle they are implanted in, as the magnets move with it.

Most of the magnetic tracking systems proposed so far reconstruct the pose of a single marker using an appropriate number of sensors. Exceptions to single marker systems are the trackers developed by Yang et al. [5], Taylor at al. [6] and Tarantino et al. [7], that considered the pose of three (15 unknowns), four (20 unknowns), and seven (35 unknowns) markers. A more recent study suggested that this limit could be overcome and that, theoretically, an indefinite high number of magnets could be properly tracked, as long as some general design rules are respected [8]. Here, we sought to transfer such findings into a more realistic scenario, and to validate them by simulating the tracking of multiple magnets implanted in an anatomically appropriate model.

Thus, using a 3D CAD anatomical model of the forearm, we simulated the implant of n magnets in n available independent muscles. Three representative amputation levels were studied, in which both n and the implant sites were defined based on the forearm anatomy and on the rules identified in our previous study [8]. Results showed that it was possible to track up to 9, 13 or 18 MMs in a proximal, middle or distal representative transradial amputations, respectively. Remarkably accurate tracking performance were achieved, as localization errors always proved below ~3% the entire trajectories of the MMs inside each muscle. These outcomes are relevant because they suggest that a large number of magnets could be implanted and effectively tracked, thus allowing to achieve independent control of multiple DoFs in a hand prosthesis.

MATERIALS AND METHODS

Three configurations resembling three possible levels of transradial amputation were simulated with the aid of a 3D CAD model of the forearm of a healthy human (50th percentile male; Zygote, American Fork, US). The first and second configurations (T1 and T2) simulated amputations occurring at the first and second proximal third of the forearm, respectively (Fig. 1). The third one accounted for an amputation across the carpal bones (wrist disarticulation) leaving most of the extrinsic hand muscles available for the implant (T3, Fig. 1). For each configuration, we first identified n, i.e. the number of magnets that could be implanted and independently tracked, and defined their position in the muscles. This was done through a placing procedure that took into account: (i) the geometry of the residual forearm; (ii) a simplified biomechanical model of the muscle contraction; (iii) general guidelines identified in our previous study [8]. We then simulated the MMs movement caused by the muscles contraction, and acquired the generated magnetic field through N simulated sensors. Finally, we ran a localization algorithm to estimate the MMs poses to verify the effectiveness of the placement procedure.

MMs were modelled as Nd-Fe-B N45 grade cylindrical magnets (axial remanent magnetization $B_r = 1.27$ T, radius = 1 mm, height = 2 mm). Only muscles that after the amputation had a residual length of at least 20% the original one [9] were considered eligible for the implant. According to such criterion, T1, T2 and T3 presented a total of 18, 19 and 23 eligible muscles, respectively (Fig. 1). The displacement of the muscles was modelled according to the following linear equation:

$$d(x) = \begin{cases} \frac{x}{L} d_{max}, & x < L \\ d_{max}, & x \geq L \end{cases} \qquad (1)$$

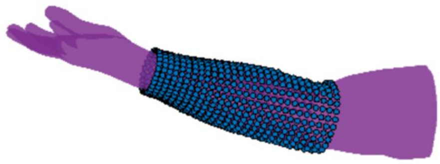

Figure 2: Sensors arrangement around the forearm, extracted from the CAD model. A grid of N = 840 sensors was used to collect the magnetic field generated by the implanted magnets. The longitudinal and radial step of the grid was set to 10 mm and 12°, respectively.

where x is the coordinate identifying the curve that runs along the muscle central axis, starting at the ideal transition between the proximal tendon and its belly; $d(x)$ indicates the actual muscle displacement; d_{max} is the maximum muscle contraction (assumed equal to 10 mm); L is the length of the muscle belly at rest.

Magnets Placing Procedure

A magnets placing procedure was implemented for defining their initial (rest) position in the residual muscles for each configuration. More specifically, we exploited an optimization procedure based on a non-linear programming solver implemented in Matlab (MathWorks, Natick, MA). Such procedure worked under the following hypotheses: (i) only one magnet could be implanted in each muscle; (ii) muscles deformation was assumed to take place only in the longitudinal direction; (iii) magnets could be implanted only in (sections of) the muscle belly, and not in the tendons; (iv) the magnetic moment vectors of the implanted MMs always pointed radially, in order to maximize the magnetic field measured by the sensors.

The procedure was initialized by placing the MMs in the center of the available muscles belly (i.e., the midpoint of the central axis). Then, it searched for a placement of the MMs that maximized both the average $d(x)$ and the average value of the geometrical parameter $\bar{R}$, defined as:

$$\bar{R} = \frac{1}{n} \sum_{i=1}^{n} \frac{L_{inter-MM_i}}{L_{MM-sensor_i}} \qquad (2)$$

where $L_{inter-MM_i}$ indicates the distance between the i-th MM and the nearest implanted MM and $L_{MM-sensor_i}$ is the distance between the i-th MM and the nearest sensor. The procedure searched for a spatial arrangement that ensured $R_i \geq 0.6$ for each magnet, a condition supposed to guarantee an accurate multi-magnet tracking [8]. Initially, a number of MMs equal to the total number of sites eligible for implantat was considered. Then, at the end of each iteration, the magnet that scored the lowest R (if below 0.6) was removed. The placement procedure was iterated until all

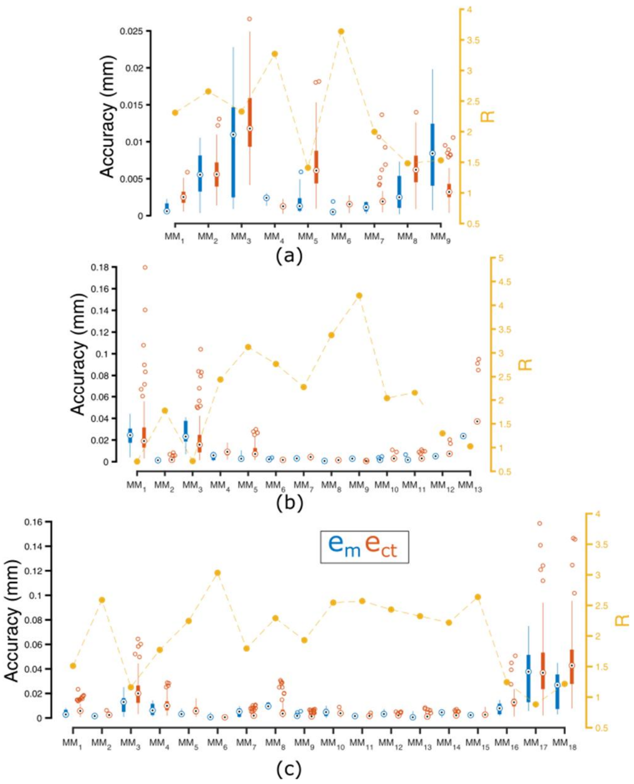

Fig. 3. Boxplots represent e_m (blue) and e_{ct} (red) for each MM, while the lines indicate the corresponding R value, for (a) T1, (b) T2 and (c) T3. In general, lower R values led to lower accuracies, and vice-versa.

remaining MMs were successfully placed (all $R_i \geq 0.6$) within the simulated forearm.

Localization Problem

Once the placing procedure was completed, the MMs were moved along the muscle axis following anatomically appropriate trajectories. Each trajectory originated in the position defined by the placing procedure, and ended proximally at a distance $d_i(x)$. The MMs displacement was approximated by translating them, one at a time, along 11 equidistant checkpoints (0%, 10%, 20%, …, and 100% the trajectory length). At each checkpoint, the analytical model described in [10] and validated in [11] was used to simulate the magnetic field generated by the MMs. Indeed, analytical approaches generally have a lower computational cost compared to numerical methods, other than being more accurate [12]. Consequently, many studies focused on the analytical calculation of the magnetic field produced by currents in a coil [13], or by arc-shaped permanent magnets (e.g. cylindrical permanent magnets) [11]. To reduce the computational burden, these analytical formulations are generally expressed in terms of complete elliptic integrals or through series expansion [14]. A compact and efficient representation of the magnetic field produced by an axially uniformly magnetized cylindrical permanent magnet was used in this work [10].

Such field was sampled on a grid of $N = 840$ simulated sensors arranged around the forearm, and ideally hosted within the prosthetic socket (Fig. 2). The longitudinal and radial step of the grid was set to 10 mm and 12°, respectively. This resulted in a distance between adjacent sensors between 6 mm and 10 mm. Sensors recordings at each checkpoint were stored and subsequently fed to a Matlab script that ran the Levenberg-Marquardt algorithm [15] to retrieve the poses of the MMs offline.

To solve the inverse problem of magnetostatics and thus retrieve the poses of the MMs, the latter were approximated as point-like dipoles, akin to our previous works [4][7][8]. The localization accuracy, both in terms of position and orientation, was evaluated as:

$$E_p \approx e_m + e_{ct} \tag{3}$$

where e_m accounts for inaccuracies in tracking the displacement of the moving magnet (i.e., model error), while e_{ct} accounts for false predictions of simultaneous displacement affecting the non-moving magnets (i.e., cross-talk effect). e_m and e_{ct} were defined as the Euclidean distance between the actual and the estimated displacement for the moving and non-moving MMs, respectively, akin to our previous works [4][7][8].

RESULTS

The placement procedure selected a total of nine target muscles for T1, 13 for T2 and 18 for T3. The minimum and maximum displacement underwent by the MMs were respectively 0.6 and 1 mm, across all configurations. For the sake of brevity we report only the results related to the position accuracy. In fact, the orientation accuracy proved always lower than 0.36° for all MMs in all configurations, and its trend closely matched that found for the position accuracy.

Regarding the localization accuracy, e_m proved always lower than 0.07 mm (Fig. 3). In particular, the highest e_m values were obtained for MM_3 in configuration T1 (e_m= 0.02 mm), for MM_{13} in configuration T2 (e_m= 0.05 mm), and MM_{17} in configuration T3 (e_m= 0.07 mm). e_{ct} proved generally higher than e_m (Fig. 3), but still in the same order of magnitude. Indeed, the highest e_{ct} values obtained were 0.03 mm for MM_3 in configuration T1, 0.18 mm for MM_1 in configuration T2, and 0.16 mm for MM_{17} in configuration T3. Overall, both e_m and e_{ct} proved lower than or equal to 3% the shortest trajectory covered by the MMs during muscle contraction (i.e., 6 mm). For both e_m and e_{ct}, the localization accuracy generally worsened when R decreased (Fig. 3). As an example, in configuration T2, MM_1 showed the highest e_{ct} (equal to 0.18; R = 0.71), while it proved always lower than 0.01 for MM_9 (R = 4.20).

DISCUSSION

In this work, we studied the effects of the complex anatomy of the human forearm on the design of a myokinetic control interface aimed at driving multiple DoFs in a hand prosthesis. A magnets placing procedure was implemented which, based on the forearm anatomy and on predefined rules, guided the choice of the number and sites of implant of multiple MMs.

Traumatic amputations lead to a large variety of muscle health conditions, which are not standardized across subjects. Our approach allows to customize the MMs arrangement based on the muscles distribution of a specific patient. The latter could be made available from 3D MRI images [16]. By enabling a preclinical planning of the MMs placement, we could significantly reduce the duration of the surgical procedure, and optimize as well as ensure good performance of the prosthesis control system.

In agreement with previous findings [7][8], localization errors generally increased for lower R values. This justifies the need for planning the MMs placement as opposed to a random one. Specifically, in [7] nine MMs were randomly distributed in a workspace that mimicked only the bulk volume of the forearm. In this case, the system failed to retrieve their poses with acceptable accuracy. Here, the imposed constraints allowed to achieve good tracking accuracies even when the number of magnets was doubled.

The present study was indeed limited in some respects. First, in order to limit the number of combinations tested, the orientation of the MMs was kept fixed (pointing towards the sensors). This is the optimal configuration, selected to maximize the magnetic field sampled by the sensors. However, variability in the orientation of the MMs is expected as these will likely be implanted manually in the muscles. Secondly, we considered a simplified linear model to describe the muscle displacement. This only captured longitudinal elongations/contractions of the muscles, while it is known that these undergo radial deformations as well. Thus, the ability of the approach in coping with more complex movements of the MMs remains to be tested.

The outcomes of this work pave the way towards the development of an intuitive control system that can be used to drive a dexterous hand prosthesis, by significantly improving both the naturalness of the control strategy and its functionality. Furthermore, they are of great interests for a multitude of bioengineering applications that exploit multi-magnet tracking in a constrained workspace.

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