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The Myokinetic Interface: implanting permanent magnets to restore the sensory-motor control loop in amputees

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Abstract

The development of a dexterous hand prosthesis which is controlled and perceived naturally by the amputee is a major challenge in biomedical engineering. Recent years have seen the rapid evolution of surgical techniques and technologies aimed at this purpose, the majority of which probes muscle electrical activity for control, and deliver electrical pulses to nerves for sensory feedback. Here we report on the *myokinetic interface* concept that exploits magnetic field principles to achieve natural control and sensory feedback of an artificial hand. Like implantable myoelectric sensors, but using passive implants, localizing magnets implanted in independent muscles could allow monitoring their contractions, and thus controlling the corresponding movements in the artificial hand, in a biomimetic, direct, independent, and parallel manner. Selectively vibrating the magnets also offers a unique opportunity to study kinesthetic percepts in humans. The myokinetic interface opens new possibilities for interfacing humans with robotic technologies in an intuitive way.

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Keywords

Magnetic sensors, magnetic tracking, myokinetic control interface, myokinetic stimulation interface, artificial hand.

1. Introduction

The restoration of dexterous motor functions and multimodal sensing equivalent to those of the human hand following amputation is one of the major challenges in biomedical engineering. The main reason for this is that amputation interrupts the bi-directional communication flow between the peripheral and the central nervous systems, with the consequent need to re-establish both the efferent and the afferent pathways. Despite all the advances brought by one century of modern research in the field – perhaps the first patent on a powered hand dates back to 1915 [1] – today's available artificial arms and hands are just rough copies of their biological model. How to decode motor volition to seamlessly control such limbs and how to provide intuitive and useful sensory feedback are – among others – still today pending questions faced by our community.

As a matter of fact, the most reliable controller available nowadays is not far from the two-state amplitude modulation electromyography controller proposed by Bottomley [2] back in the '60s, in which a single pair of antagonistic muscles controls the opening and closing of the prosthetic hand. Despite its intuitiveness and ease of fitting, this scheme cannot differentiate between different muscular patterns pertaining to different hand movements, and, accordingly, cannot be used to control multiple grasps of a dexterous prosthesis. Even more, in spite of decades of research efforts for the search of useful feedback strategies, sensory feedback systems are still not clinically available [3]. All of this corroborates the need to pool past and current efforts to make effective solutions readily available to the end-users.

In this context, a new portfolio of possibilities recently emerged from the union between technology and surgery, aimed at restoring a more natural control and sensory feedback in prosthetic limbs [4–6] (Figure 1). The expression *bionic reconstruction* is increasingly used in the scientific community to denote such recent advances. Bionic reconstruction appears to be a viable, and in some respects advantageous alternative to hand transplantation. In fact, while remaining a valuable solution for bilateral amputees and providing a real "like with like" substitution, transplantation comes with significant drawbacks like the need for lifelong immunosuppressive therapy, which make the bionic alternative attractive ([6,7]). Finally, bionic reconstruction was also proposed as a viable solution following elective amputation, to get rid of a functionless and potentially painful hand [8].

Among the surgical techniques for improving motor control, Targeted Muscle Reinnervation (TMR) has certainly gained worldwide impact in the past two decades [9]. Proposed by Hoffer and Loeb in the 80s [10] and brought to clinical reality by Kuiken in 2002 [9], TMR concerns the grafting of nerves formerly innervating the missing limb into remaining muscles in the stump region, which behave as selective biological amplifiers of the nerve activity. The muscle electrical activity can be probed with electrodes and provides an excellent target for control signal acquisition. Following the same concept, peripheral nerve bioamplifiers can be created by

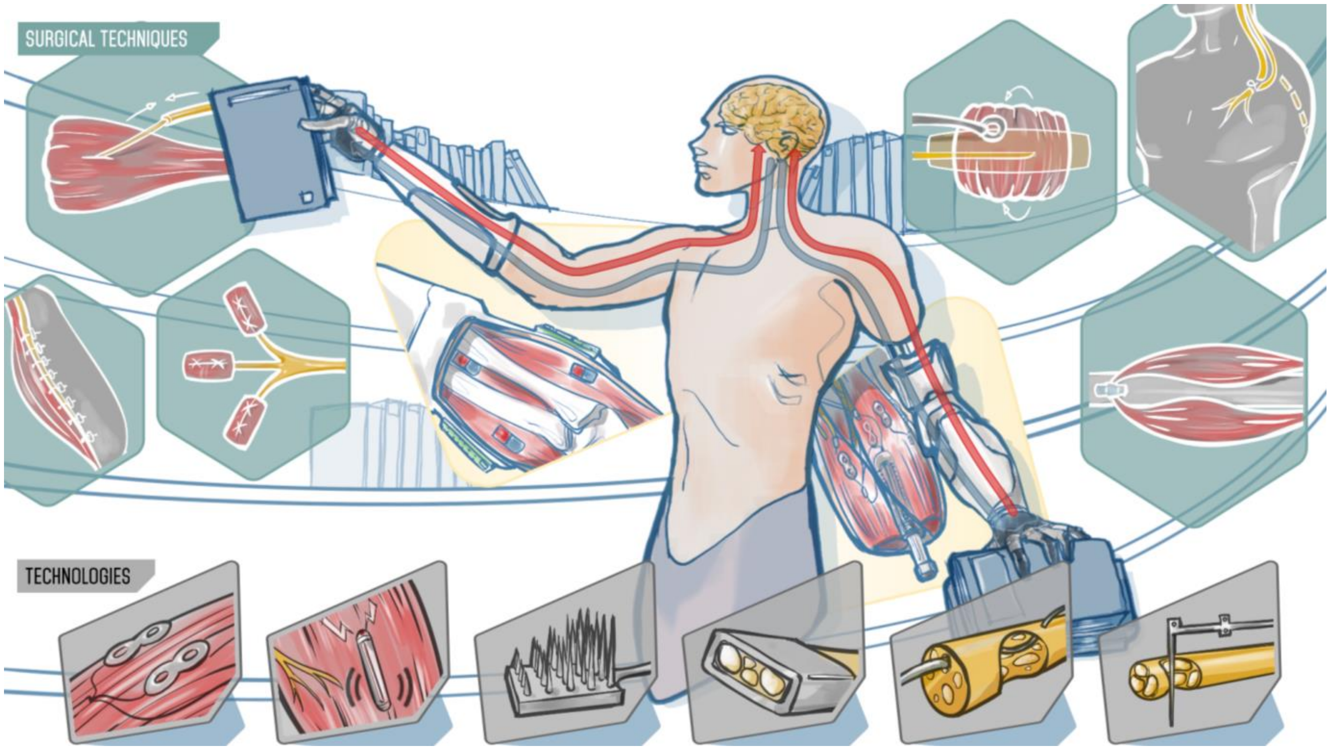

Figure 1. State-of-the-art surgical techniques and technologies for the sensory-motor control loop restoration in upper limb amputees. Upper left corner: surgical approaches to access motor neural signals using muscles as biological amplifiers; TMR (upper inset) [9] and RPNIs (lower right inset) [11] create novel sources of control; composite RPNIs include skin grafts for sensory feedback (C-RPNIs, lower left inset) [28]. Lower left corner: epimysial electrodes (left) [13] and IMESs (right) [15] can be used to acquire selective control signals from muscles to achieve control. Upper right corner: surgical approaches to enable sensory perception from missing limbs. In TSR (upper right inset) [24] and cutaneous mechanoneural interfaces (upper left inset) [29] the severed sensory nerves are transferred to denervated skin patches in the stump or skin grafts, respectively. AMI (lower inset) [32] allows the restoration of proprioceptive feedback by coapting antagonistic pairs of muscles. Lower right corner: (from right to left) intra-neural (TIME, LIFE), extra-neural (FINE), and penetrating (USEA) neural electrodes [22] interface directly with the peripheral nervous system at different levels of invasiveness, to restore close-to-natural sensory feedback through electrical pulses. Central-left inset: the myokinetic interface foresees the implantation of permament magnets to monitor muscle displacement and restore proprioceptive sensations [43]. Central-right inset: the neuromusculoskeletal interface, proposed by Brånemark and Ortiz-Catalan, is the first chronically implanted bi-directional interface brought to clinical reality [12].

implanting the transected peripheral nerves into muscle grafts, a solution proposed by Cederna and colleagues and named Regenerative Peripheral Nerve Interface (RPNI) [11]. Other than creating more control sources, efforts in surgery have been devoted to improving the mechanic attachment of prosthetic devices to the human body, finding their main expression in osseointegrated prostheses [12]. Featuring a direct attachment of the end effector to the bones through percutaneous titanium implants, the latter replace conventional suspension systems and offer higher stability, range of movement and comfort. Brånemark and Ortiz-Catalan exploited the access provided by the percutaneous implants to demonstrate the first permanent neuromusculoskeletal prostheses in transhumeral and transradial amputees [12–14]. Epimysial and cuff electrodes interfaced with the muscles and nerves and wired through the implants, allowed for control, and neurostimulation, respectively. In parallel, wireless implantable myoelectric sensors (IMESs), proposed by Weir in 2009 [15] and first demonstrated in a human subject by Pasquina in 2015 [16], proved as the first sensing technology to obviate the use of wires (Figure 1).

The purpose of TMR and implantable recording technologies is that of creating and accessing multiple, independent control sources characterized by high sensitivity, to achieve direct and simultaneous control over multiple DoFs. However, obtaining independent control signals is not trivial, as reinnervated muscles contain rich neural information corresponding to all muscles of the lost limb, which is difficult to disentangle. A solution to this problem is offered by machine learning or pattern recognition, a technique extensively investigated for more than 50 years [17], although becoming clinically available in the early 2000's [18–20]. The vision is that by combining the benefits of complex bioamplified control signals, stable mechanical attachment, selective recording technologies, and advanced control algorithms should lead the way towards a multi-DoFs control of multi-articulating prosthetic limbs in a more natural way.

However, restoring control alone might not suffice: grasping and manipulation heavily rely on tactile information, therefore it appears reasonable that prostheses would perform better if used in a closed-loop with the user [21,22]. To achieve this goal, the prosthesis should not only be able to detect physical interactions with

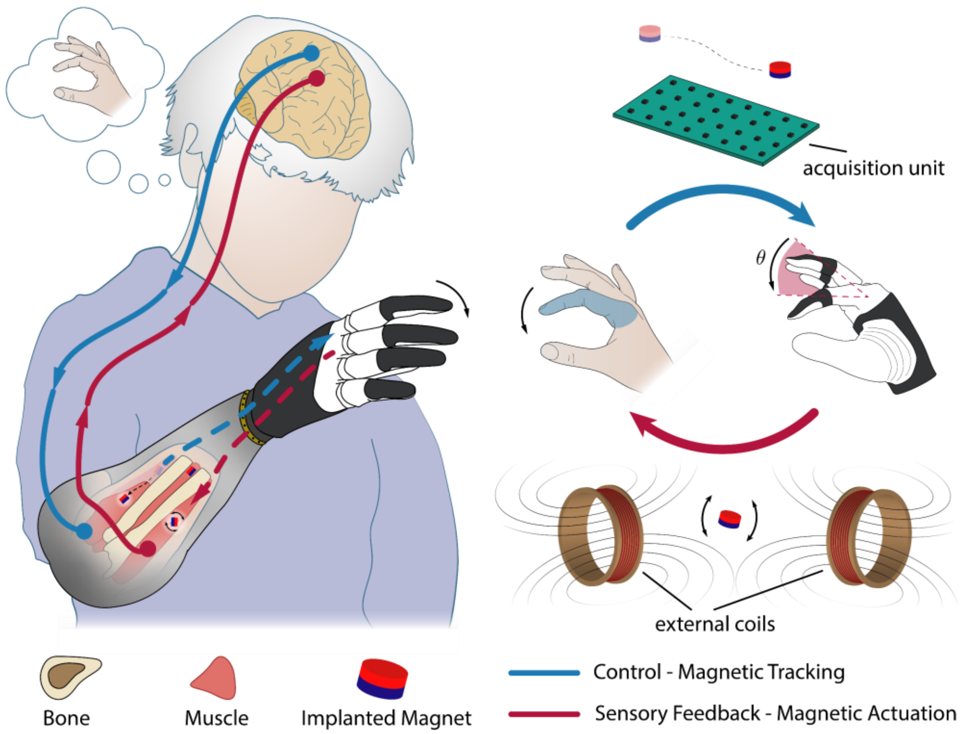

Figure 2. Overview of the myokinetic interface. Permanent magnets are implanted in relevant wrist muscles and extrinsic muscles of the hand. Signals generated by the magnets during voluntary contraction are sensed by external acquisition units, and used to estimate their pose. This information can be used to control the physiologically appropriate degree of freedom of the prosthetic wrist and hand in a direct and proportional manner. At the same time, a selective vibration can be induced in the corresponding magnet using external coils, to activate receptors responsible for proprioceptive sensations. In this way, the user could be made aware of the activated joint position in space, with no need of looking at the prosthesis.

the environment, and sense its internal state, but also be able to convey such information to the user in a perceivable and possibly effortless manner. Over the years, various surgical techniques and implantable technologies have been proposed to deliver effective feedback in amputees (Figure 1). Starting with surgical techniques, an exciting solution was discovered when the first patients who received TMR reported about hand sensations in the skin area overlying the reinnervated muscles [23]. This idea was further investigated by deliberately denervating the cutaneous nerve branches, and coapting them with reinnervating main nerve trunks from the stump [24]. Hebert and colleagues demonstrated that such a technique, called Targeted Sensory Reinnervation (TSR), could restore multiple sensory modalities via non-invasive skin stimulation (touch, temperature, pain) [24]. In addition, studies with TSR patients conducted by Marasco's group demonstrated improvements over the control of a robotic arm via non-invasive stimulation to provide kinaesthetic percepts and/or touch [25], [26]. Recently, research groups expanded the potential of RPNIs, including sensory feedback. Cederna and colleagues showed that direct electrical stimulation of RPNIs could produce tactile and proprioceptive sensations [27], paving the way for composite structures integrating skin grafts for an increased sensitivity [28],[29].

Reliving and renovating the ancient idea of cineplasty [30,31], Herr's group proposed the agonist-antagonist myoneural interface (AMI) to activate receptors responsible for proprioceptive sensations during motion [32]. This becomes possible by creating natural agonist/antagonist muscle pairings through surgical tendon connections, and to date it was demonstrated in lower limb (below-knee) amputees. It remains unclear whether the AMI may be adapted to the upper limb, for below-elbow amputations, where complex relationships between agonist-antagonist muscles for finger and wrist control do exist [26].

Since action potentials travel in the form of electrical signals, in principle it could be possible to convey somatosensory percepts by electrically stimulating the nerves originally serving the digits and palm, using neural electrodes [22]. Cuff, TIME, LIFE, FINE and USEA are among the types that have been used to provide *close-to-natural* tactile sensations [33–35], and most of all have demonstrated stability with time, after the pioneering work by Clippinger and Reswick in the 70s [36,37]. Neural sensory feedback proved clinically viable for long term home use in the neuromusculoskeletal prosthesis [12,13], and, in certain cases, to improve motor control [38–41] However its feasibility to actually restore natural sensations is still a controversial matter [42].

Within this exciting and rapidly evolving scenario, we proposed an alternative to control and feedback methods based on neuromuscular electrical signals, that we called the *myokinetic interface* (Figure 2). This solution deploys permanent magnets implanted with a minimally invasive procedure in the residual muscles, which are localized/tracked to derive control commands associated to muscle contractions [43]. Briefly, the magnetic field produced by the magnets is recorded by sensors placed around the residual limb on the suspension system (e.g. the prosthetic socket), and processed to independently track the movements of the magnets. This information, correlated to motor intention, could thus be used to command the end effector, e.g. the closing/opening movements of the prosthetic hand. As an example, magnets implanted in the residual thumb flexor muscle (*flexor pollicis longus*) could be used to control the flexion of the artificial thumb in the prosthetic hand. More in general, localizing magnets implanted in multiple muscles could allow monitoring their contractions, and thus controlling the corresponding movements in the artificial hand, in a *biomimetic*, *direct, independent,* and *parallel* manner. Not only, having magnets implanted in the muscles offers an exceptional window for investigating the kinaesthetic sensation, by selectively activating proprioceptive receptors through remote vibration (Figure 2).

The myokinetic interface finds its highest clinical and scientific motivation in treating trans-radial amputations, as magnets implanted in the several extrinsic muscles of the forearm could potentially restore the stunning dexterity of hand movements. Nevertheless, the same approach could be easily extended to and combined with other amputations levels or surgeries (e.g.: glenohumeral amputations with TMR, transtibial ones treated with AMI, partial hand amputations, etc.) or to treat diseases requiring movement assistance or rehabilitation (e.g.: exoskeletons).

In the past years, we have investigated both the technical feasibility and the scientific questions raised by this novel approach, as described in this paper.

2. The Myokinetic Interface

The myokinetic interface comprises of magnets implanted in independent muscles, and external *magnetic tracking* and *actuation* systems capable of: (i) continuously localizing the movements of the magnets and, at specific times, (ii) inducing subtle movements in specific ones. The purpose of this architecture is to restore both the *motor control* and *sensory feedback paths*, respectively, in the individual receiving the implants (Figure 2). More in detail, the magnetic tracking system monitors the physical displacements associated to contractions in different muscles and uses this information to send control commands to the prosthetic hand. In parallel, the magnetic actuation system induces vibrations in the magnets to convey kinaesthetic sensations (by activating muscle spindles and Golgi tendon organs) in response to the sensory status of the prosthetic hand. Notably, passive implants (meaning that they do not require batteries or power sources) and wearable devices could make for a bidirectional human-machine interface (HMI) with enhanced capabilities with respect to the state of the art. As the envisioned HMI would transduce residual muscle movements into decipherable signals, we borrowed from the Greek roots and called it *MyoKinetic interface*.

2.1 The control path: magnetic tracking

Muscle contraction can be monitored by implanting and tracking the pose (position and orientation) of one or multiple magnets per muscle. The tracking system is responsible for continuously retrieving such poses, using magnetic field sensors and solving the so called magnetic inverse problem [43]. In brief, the theoretical magnetic field B_i generated by a single magnet at the position of the i^{th} sensor, is modelled after the vectorial equation of the magnetic dipole:

$$\boldsymbol{B}_i = \boldsymbol{B}(x_i) = \frac{M\mu_r\mu_0}{4\pi}\left(\frac{3(\widehat{\boldsymbol{m}}\cdot\boldsymbol{x}_i)\boldsymbol{x}_i}{|\boldsymbol{x}_i|^5} - \frac{\widehat{\boldsymbol{m}}}{|\boldsymbol{x}_i|^3}\right) \tag{1}$$

where M and m are the magnitude and direction of the magnetic moment of the magnet, respectively; x_i is the vector distance between the magnet and the i^{th} sensor.

In the case of N sensors and n magnets, $\boldsymbol{B}_i$ at the location of the i^{th} sensor can be modelled as the linear superimposition of that generated by each dipole. Thus, for each i^{th} sensor, the following equation applies:

$$\boldsymbol{B}_i = \boldsymbol{B}(x_i) = \sum_{j=1}^{n}\frac{M_j\mu_r\mu_0}{4\pi}\left(\frac{3(\widehat{\boldsymbol{m}_j}\cdot\boldsymbol{x}_{ij})\boldsymbol{x}_{ij}}{|\boldsymbol{x}_{ij}|^5} - \frac{\widehat{\boldsymbol{m}_j}}{|\boldsymbol{x}_{ij}|^3}\right), \text{i=1,…,N} \tag{2}$$

where j indicates the j^{th} magnet. Reversing Equation (2) provides the absolute pose of n magnets which can be computed through numerical approximation methods or solvers [43]. This is a mathematical problem with 5 unknowns for each magnet (3 for position and 2 for orientation) and requires the measurements from at least two 3-axis sensors. Yet, as numerical solvers are typically more accurate when the number of equations is much larger than the number of unknowns, the tracking accuracy improves with the number of sensors [44,45]. The use of numerical solvers and the magnetic dipole model introduce errors in estimating the pose of single magnets (model error - e_m), and can yield to false estimates of simultaneous movements (cross-talk – e_{ct}) in the case of multiple magnets (see Tarantino et al. for a detailed mathematical description [43]). In addition, technological limitations (sensor resolution and sampling frequency) and environmental factors (magnetic interferences or noise) may further degrade both the tracking accuracy and precision (repeatability).

Once the magnet poses are available, the degrees of muscle contraction can be inferred either from the displacements of the magnets from offset poses (recorded with uncontracted, relaxed muscles), in the case of a single magnet [43], or from the relative distances between paired magnets, in the case of multiple magnets [46,47]. In all cases, for an effective operation, the magnets should be implanted in sites that maximize the physical displacement range during contraction.

2.1.2 What we learnt

In the past years we comprehensively searched for general design rules for a magnetic tracking system,

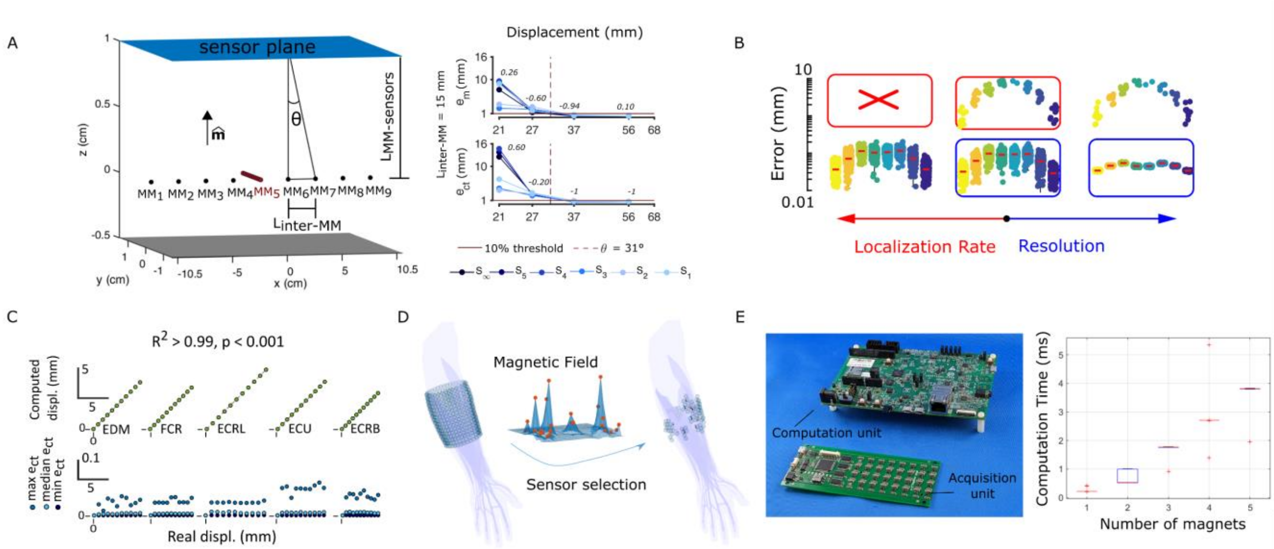

Figure 3. The control path: magnetic tracking. (A) θ was identified as a descriptive parameter of the system geometry. For θ values higher than 31°, the model error e_m and the cross-talk error e_{ct} proved below 10% the trajectory length covered by the central magnet (MM_5) for different sensing surfaces (S_1-S_5, S_∞). [48]. (B) Localization error is affected more by localization rate than by sensor resolution [49]. (C) Relative and computed displacement (displ.), and localization error of five out of eleven magnets virtually implanted in different muscles of the forearm (e.g. EDM – Extensor Digiti Minimi) in a representative proximal amputation. e_m and e_{ct} proved always below 0.67mm and 0.30mm, respectively. Acronyms in [51]. (D) An optimal sensor set (blue dots) can be derived based on the peaks (orange dots) generated by the magnets on the magnetic field and its gradient (not shown) [52]. (E) The first prototype of embedded tracking system consisted of an acquisition unit hosting 32 sensors and a computation unit that implemented the tracking algorithm. The computation time increased with the number of magnets and proved always below 4ms [53].

capable to serve as a controller to prosthetic limbs or other assistive devices. Our results corroborated early findings, i.e. that tracking errors increase with the number of magnets and the distance to the sensors, and decrease with the number of sensors [44,45]. Yet, we contributed to knowledge by investigating multi-magnet tracking systems capable to localize many more magnets than those described earlier (mostly single magnet systems).

In Gherardini et al. [48], we systematically analyzed the effects of remanent magnetization, number of sensors, and geometrical configuration (i.e. distance among magnets: $L_{inter\text{-}MM}$, and between magnets and sensors: $L_{MM\text{-}sensor}$) on the tracking accuracy. We simulated (and experimentally validated) a general yet realistic setup, including a sensing plane with uniformly distributed sensors, $L_{MM\text{-}sensor}$ far from remote magnets (Figure 3A). We found that the tracking accuracy is mainly affected by the specific angle $\theta = \tan^{-1}(L_{inter\text{-}MM} / L_{MM\text{-}sensor})$, and that if θ is greater than ~31°, and enough sensors are available, *an indefinitely high number of magnets can be accurately tracked*. In Masiero et al. [49], instead, we simulated the effects of the intrinsic properties of the sensors on the accuracy and computation time of the numerical solver, during muscle contraction. We found that *the accuracy is primarily and positively affected by the localization rate* (which is directly related to the sampling frequency), and less affected by the sensor resolution (Figure 3B). Accordingly, *the computation time (or number of iterations of the solver to converge) decreases with the localization rate*.

In studies more tailored on the final application we validated these rules using a physical forearm mockup and a finite element model of the forearm muscles, to determine the maximum number of magnets that could be properly localized in such workspaces [43,50–52]. In Milici et al. [51], we found that by respecting the "θ rule", up to 11, 13 and 19 magnets can be implanted in representative proximal, middle and distal amputation levels, respectively. Most of all, these magnets can be tracked with minimal localization errors (below 7% the trajectory travelled by the magnets during muscle contraction) (Figure 3C). In another study [52], concerned about the number of sensors that would be needed in reality (and on the acquisition time to retrieve such readouts), we sought to identify strategies to reduce them. We proposed the *Peaks* method: a computationally inexpensive strategy to define and select the minimum set of sensors for tracking multiple magnets in an accurate way (Figure 3D). As per this approach, only the sensors capturing the peaks of the magnetic field and its gradient are selected over the initial sensor grid, using a simple thresholding technique. This demonstrated capable to select <20% of the initial sensor set (from 480 to 80), yet ensuring an accuracy statistically comparable to the initial set, and to more complex, state-of-the-art methods.

Besides these studies, we developed two magnetic tracking systems prototypes suitable for integration on wearable devices [53,54]. In Clemente et al. [53], we proved for the first time the viability of using an embedded system for real-time magnet localization (Figure 3E). A grid of 32 sensors and a microprocessor-based computation unit were used to retrieve the poses of multiple magnets in real time, running the Levenberg-Marquardt algorithm (LMA). The device proved capable of localizing up to five magnets (>250 Hz rate) attached to artificial muscles in a physical forearm mockup, replicating the extrinsic muscles of the hand. It demonstrated highly precise (1% repeatability) yet

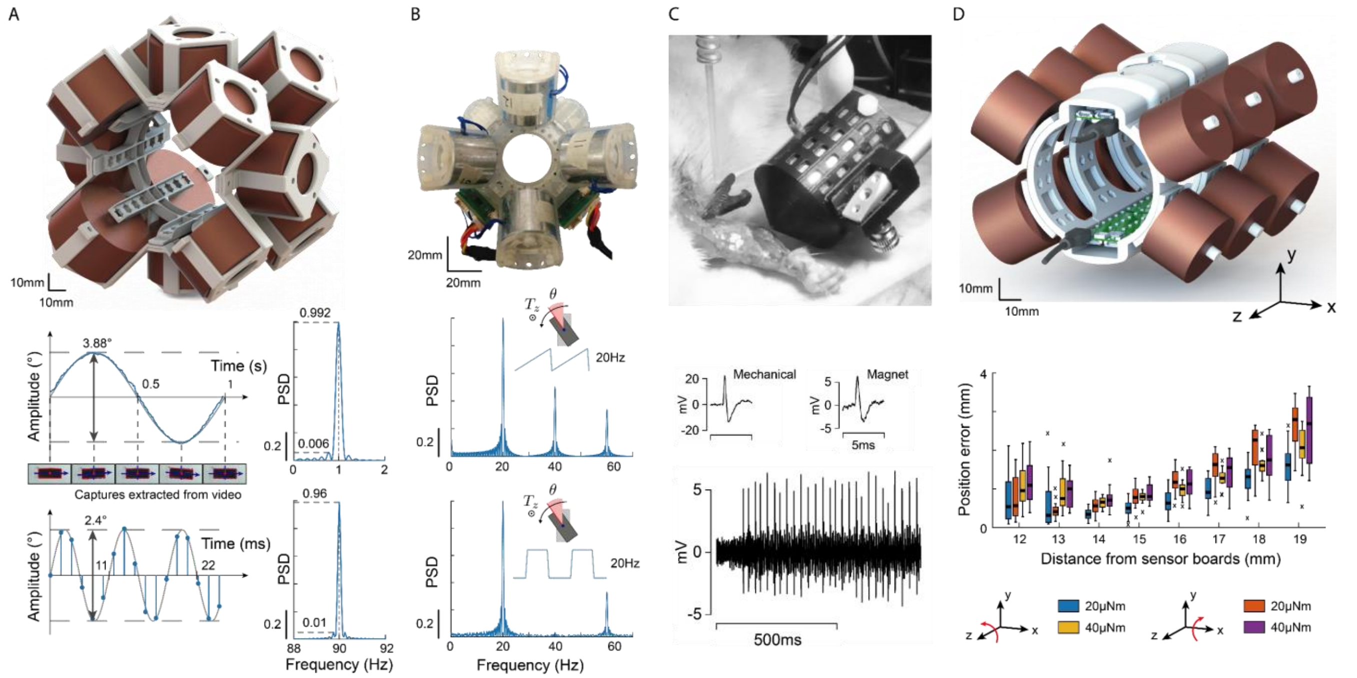

Figure 4. The feedback path: magnetic actuation. (A) First prototype of stimulation system consisting of 12 coils [58] (upper panel); magnet displacements and Power Spectral Densities (PSDs) following sinusoidal 1Hz and 90Hz torsional vibrations induced by the system (lower panel). (B) Second prototype of stimulation system consisting of eight coils [56] (upper panel); PSD of 20Hz sawtooth and squarewave torsional vibrations induced with the system (lower panel). (C) In vivo experiments on rodents to evaluate the feasibility of inducing muscle-sensory responses through magnet vibrations [56] (upper panel); direct comparison between action potential profiles of a single unit muscle sensory receptor induced by the mechanical system ('Mechanical') or by the linear permanent magnet system ('Magnet'), and representative single unit neural responses to 100Hz sinusoidal displacements induced by the linear permanent magnet system (lower panel). (D) Last prototype of stimulation system with 12 coils and 4 magnetic sensor boards for simultaneous tracking and actuation (upper panel); tracking accuracy of 1 magnet vs. distance from the sensor boards achieved with the system (lower panel).

exhibiting limited cross talk errors corresponding to 10% the mean trajectory undergone by the magnets. Compared to a standard PC implementation, it exhibited similar precision and accuracy, while being ~75% faster. In Pertuz et al. [54], we assessed the use of machine learning models in place of LMA for localizing a single magnet implanted in the mockup, using 128 sensors. Two data-driven models were implemented on field-programmable gate arrays using customized floating-point operators. The system was tested offline and demonstrated a tracking accuracy of 720μm and a computation time of 12.07μs.

In [55] we demonstrated the combination of the TMR procedure with the myokinetic interface, in a non-invasive way, with two individuals who underwent TMR following above-the-elbow amputation. We assessed the feasibility of discriminating three different DoFs (six movements) of the missing limb by tracking the displacement of five magnets placed on the skin, over the reinnervated sites. A simple logistic regressor proved able to discriminate the different DoFs, with an average F1-score among classes and testing conditions of 0.69 and 0.60 for the two participants, respectively.

Finally, in [47] we compared the tracking accuracy achieved by implanting a single magnet per muscle, versus implanting pairs of magnets per muscle, under ideal and mechanically disturbed conditions (i.e. shift of the sensor grid). We found that implanting one magnet per muscle always lead to lower tracking errors under ideal conditions. However, as expected, when mechanical disturbances were applied, magnet pairs proved capable to reject the disturbances, while the single magnet approach could not.

Overall, these findings contributed providing important guidelines for the design of a myokinetic control interface, and more in general, for a wider range of biomedical applications exploiting magnetic tracking.

2.2 The sensory feedback path: magnetic actuation

The myokinetic interface presents itself as an unprecedented scientific instrument and method to study proprioception in humans. Indeed, magnets implanted in the muscles can be remotely vibrated (in the 70–115Hz range [25]) using wearable coils so as to stimulate purely kinaesthetic responses that are fully decoupled by skin receptors [56]. To this aim the actuation system of the myokinetic HMI comprises electromagnets (or coils) distributed around the residual limb, and a current controller. The latter computes and regulates the electrical currents flowing in the coils, to produce the magnetic field in time and space, that induce the desired interactions with the magnets, to vibrate them (Figure 2). These currents are computed by reversing the equations describing the magnetic interaction between C sources (coils) and n targets (magnets), by imposing the desired force $\boldsymbol{F}$ and torque $\boldsymbol{T}$ vibration patterns:

$$\boldsymbol{F} = \nabla(\boldsymbol{B}(\boldsymbol{x}_i) \cdot \boldsymbol{m}_i), \ \ \boldsymbol{T} = \boldsymbol{m}_i \times \boldsymbol{B}(\boldsymbol{x}_i) \tag{3}$$

where $\boldsymbol{B}$ is the compound magnetic field generated by the sources at the target locations $\boldsymbol{x}_i$, and is proportional to the currents, whereas $\boldsymbol{m}_i$ are the target orientations (see [57] and [58] for a comprehensive description).

As per this problem, in principle, the complete control of n target magnets requires $8n$ sources (3 for the torque and 5 for the force) [59]. However, as in our application magnets are not suspended in the air but constrained in an elastic medium (muscle tissue), and should only vibrate, full controllability can still be achieved with less sources and isotropic workspaces [56,57].

2.2.2 What we learnt

To date several magnetic actuation prototypes were developed to prove the feasibility of delivering: (i) finely tuned and selective vibrations to the magnets [58], and (ii) kinesthetic percepts in-vivo [56].

In Montero et al. [58] we demonstrated a real-time 12-coil prototype able to induce selective and highly directional sinusoidal vibrations in magnets, in a material resembling the viscoelastic properties of the muscles. The system proved able to update currents at a 500Hz rate, and to selectively vibrate one out of four magnets with linear and torsional vibrations, at low (1Hz) and high frequencies (90Hz) (Figure 4A). Purely torque vibrations were controlled with higher efficiency, having ~90% power spectral density located at the desired frequency (Figure 4A). The vibration amplitudes proved around 2.5-4° and up to 150μm for torsional and linear vibrations, respectively. A second prototype with 8 coils was developed to further assess the system functionality, focusing on the control of a single magnet, with different stimuli (Figure 4B) [56]. The device proved able to impose sinusoid, square wave, and saw-tooth linear and torsional vibration patterns, at both low (20Hz) and high frequencies (90Hz) (Figure 4B). However, the directionality and selectivity of the vibrations demonstrated to be largely influenced by the geometry of the workspace, and poorly influenced by far-away coils [56]. In a proof-of-concept study, we demonstrated a myokinetic actuation system feasible for eliciting muscle-sensory responses in an animal model across multiple frequencies, including those that activate kinaesthetic percepts (Figure 4C) [56].

To finely control the vibrations of magnets displaced due to muscle contraction, or in other words, to regulate the magnetic field interacting with the magnets, the knowledge of their poses is fundamental. While the most straightforward choice would be to combine magnetic tracking and magnetic actuation, such solution is technically challenging. In fact, it implies sensors sensitive to subtle magnetic fields produced by the magnets but neither saturating nor exhibiting hysteresis under the large fields produced by the coils and required for the actuation [60]. We faced this challenge by developing a system able to filter out from the sensor readouts the field contributions produced by the coils (Figure 4D). The system demonstrated capability of tracking up to 4 magnets (median position error less than 1mm) while simultaneously vibrating them with 90Hz torsional vibrations (above 80% of efficiency), keeping their vibration amplitude constant during motion.

Taken together, these findings suggest that magnetic actuation holds the promise to induce vivid kinesthetic percepts using micron-level displacements, fully decoupled by cutaneous sensations.

2.3 Implantable magnets

NdFeB magnets likely represent the best option for a myokinetic interface. As they are featured by high coercivity, high remanence, and high volume to force density, they represent an excellent compromise between high force and miniaturization. However, biocompatibility of NdFeB magnets represents a significant issue when direct implant is pursued. Hence, mechanical, in vitro, and animal tests were carried out to identify a suitable biocompatible material for coating/encapsulating the magnets [61]. First, in vitro tests investigated the mechanical resistance, corrosion resistance, and cytotoxicity relative to three different coatings, namely Gold, Titanium Nitride, and Parylene C, whose adhesion to NdFeB magnets and biocompatibility was already assessed in other industrial and biomedical applications. In vitro testing in a tissue-mimicking environment and upon contact with C2C12 myoblasts enabled assessment of the superiority of Parylene C coated magnets in terms of corrosion prevention and lack of cytotoxicity. Thus, Parylene C was tested in vivo, by implanting coated magnets in rabbit muscles for 28 days. The test showed lack of irritation and toxicity associated with the implant, thus confirming its safety.

Building on this, Taylor et al. [62] demonstrated the biocompatibility of Parylene C coated magnets for longer intramuscular implantation periods (26-weeks) in rabbits. Specifically, they used a multi-layer coating including a layer of nickel-copper-nickel, one of gold, and an outer coating of Parylene C. Again, this coating proved non-irritant, thus indicating the viability of employing Parylene C coated magnets for intramuscular implantation safely, also for long-term implantation periods.

3. Other Investigators

The emerging idea of the myokinetic interface has started producing some impact in the community, as demonstrated by the rise of scientific papers and projects embracing the approach.

The most striking one is certainly that of Moradi and colleagues [63], which recently reported the first implementation of a myokinetic control interface. They implanted single magnets in three flexor muscles (flexor digitorum profundus, flexor carpi radialis and flexor digitorum superficialis) of a trans-radial amputee, demonstrating the clinical viability of the approach. It should be noted that, while the authors presented for the first time the surgical implementation of a myokinetic interface, associated with a tendon transfer technique to increase the muscle range of motion, important limitations were present regarding the applied control algorithms.

Indeed, only a simple controller exploiting one magnet and one sensor to recognize one gesture was assessed online, while more complex signals could only be processed offline due to computational time issues.

As anticipated above, Taylor et al. [46,64] also contributed to the idea and proposed the use of pairs of magnets per muscle, to measure in-vivo tissue length. They proved real-time muscle length tracking of a turkey's gastrocnemius, by monitoring the relative distance between the implanted magnets. While, on the one hand, the use of two magnets per muscle could provide an effective solution against mechanical disturbances acting on the socket, on the other hand, it inevitably increases the computation cost and instability of the numerical solver [47,49], as well as the complexity/invasiveness of the surgical procedure [51].

4. Conclusions and Future Perspectives

The myokinetic interface holds the potential to restore natural control and sensory feedback in upper limb amputees, both at the trans-radial level (as initially conceived) and at higher levels of amputation. Such interface could be readily integrated with advanced surgical techniques, like TMR and osseointegrated prostheses, to achieve the longed-for bionic reconstruction which synthetises the recent efforts in surgery and technology. Looking ahead, many other applications in the rehabilitation field could benefit from the introduction of the myokinetic interface, among which we cite the control of lower limb prostheses as well as that of assistive exoskeletons. In a broader context, implanted magnets could be employed in human augmentation applications, to enable the remote control of different devices.

To assess the system feasibility and potentialities, we anticipate forthcoming clinical trials in amputees to test its applicability for both short- and long-term periods. This will require further assessments, especially in terms of implant biocompatibility and optimal choice and design of the electronic equipment. On top of that, future studies should be devoted to bringing valuable contributions to the scientific debate on the understanding of the basic principles, and of the cognitive processing of proprioception in humans. Nonetheless, although magnetic actuation represents an interesting tool to study proprioception, so far, its deployment in a portable device still poses significant technical challenges, due to the weight and power consumption of the electromagnetic components.

To conclude, we do hope that the development of the myokinetic interface will pave the way towards a new generation of bionic limbs and assistive devices, potentially allowing the restoration of the natural (or close-to-natural) sensory-motor control loop.

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Declaration of interests

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