A Miniature Vibrotactile Sensory Substitution Device for Multi-fingered Hand Prosthetics

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The DOI of the final edited version of this paper is: 10.1109/TBME.2011.2173342

Manuscript accepted for publication on IEEE TBME, 2011

A Miniature Vibrotactile Sensory Substitution Device for Multi-fingered Hand Prosthetics

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***Abstract*— A multi-site, vibrotactile sensory substitution system, that could be used in conjunction with artificial touch sensors in multi-fingered prostheses, to deliver sensory feedback to upper limb amputees is presented. The system is based on a low cost/power/size smart architecture of off-the-shelf miniaturized vibration motors; the main novelty is that is able to generate stimuli where both vibration amplitude and frequency as well as beat interference can be modulated. This work is aimed at evaluating this system by investigating the capability of healthy volunteers to perceive –on their forearms– vibrations with different amplitudes and/or frequencies. In addition, the ability of subjects in spatially discriminating stimulations on 3 forearm sites and recognizing 6 different combinations of stimulations was also addressed. Results demonstrate that subjects were able to discriminate different force amplitudes exerted by the device (accuracies greater than 75%); when both amplitude and frequency were simultaneously varied, the pure discrimination of amplitude/ frequency variation was affected by the variation of the other. Subjects were also able to discriminate with an accuracy of 93% three different sites and with an accuracy of 78% six different stimulation patterns**[·]

***Index Terms*— artificial hands; haptic perception; sensory substitution; vibrotactile feedback; upper limb prosthetics.**

I. INTRODUCTION

THE amputation of the hand causes severe impairments in subjects in the ability to carry out activities in daily life and in sensing the surrounding environment through one of the most important sensory organs. In modern myoelectric prostheses, although an acceptable level of dexterity is often restored by means of motorized components (e.g. hand, wrist, elbow) and electromyographic control [1], afferent sensory biofeedback is still not purposely provided. Myoelectric prosthesis users often rely on indirect feedback obtained from the sound and vibration of the motor(s) or from the kinaesthetic reaction of the socket to control their artificial limb, albeit surveys report they would like to get enhanced feedback from it [2], [3]. The lack of sensory feedback is a drawback indeed, and one of the main reasons for rejection of a prosthesis [4]. An interesting historical prospective of this problem was already presented by Childress back in the 80's [5], demonstrating the long-standing interest in this topic by scientists and engineers.

In theory, there are two ways to 'replace' the afferent pathway in amputees: (i) invasively, by interfacing directly to neural structures normally involved in the control (like the peripheral nerves [6], [7]) or (ii) non-invasively, by providing feedback to intact sensory systems normally not involved in the task (e.g., tactile stimuli on the stump). In both cases the subject should be trained to associate stimuli to events occurring to the artificial hand and fingertips. Among the non-invasive methods, the most investigated have been electrocutaneous, vibrotactile, or force/pressure feedback systems [8]-[16], i.e. devices able to display a unique modality of physical stimulation. In addition, recently Kim et al. [17] developed and successfully investigated a multi-function haptic device able to display touch, pressure, vibration and shear force on transhumeral amputees that underwent targeted muscle reinnervation [18]. This device is potentially interesting, although it is complex and could be cumbersome if applied to the forearm of trans-radial amputees.

Vibrotactile stimulation is evoked by a mechanical vibration of the skin, typically at frequencies ranging between 10–500 Hz [8]. Afferent biofeedback based on this principle has been investigated during the last decades due to the non-invasive nature of such approach that promises higher acceptability compared to e.g. electrocutaneous stimulation [8], [10]. Additionally, vibrotactile systems are small and unobtrusive allowing easier integration with respect to force/pressure feedback systems [10], [11]. The two main features of a vibrotactile stimulus are vibration **amplitude** and **frequency**, and their perception by humans on different body parts has been studied in depth. The amplitude discrimination threshold (i.e., the lowest perceivable difference in vibration amplitude) for the hairy skin of humans is known to be approximately an order of magnitude higher than for glabrous skin [19]. This threshold depends on the amplitude of the reference stimulus [20] and varies with the stimulation frequency: in particular, for the hairy skin it decreases at higher frequencies [21]. The ability to discriminate changes in frequency depends on the given stimulation frequency [21]-[22], and best results are obtained when this is above 200 Hz. Previous studies also reveal that the frequency discriminative performance is similar in hairy and glabrous skin, especially at frequencies above 50 Hz [21]. Studies relative to the cross-effect of amplitude on the

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Manuscript received August 1, 2011. This work was supported by the Ministero dell'Istruzione, dell'Università e della Ricerca, under the PRIN 2008 OPENHAND Project prot. PMZT2Z.

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frequency perception were also carried out [23].

Due to the relative ease of use, low cost and size, vibrotactile systems using small electric motors have been used in research with the Otto Bock and Motion Control myoelectric prostheses [24], [6], [14] the Cyber- [25], MANUS- [26] and the Fluid- hands [27].

This paper presents a flexible vibrotactile system composed of **vibr**atory **el**ements (hereafter named as *vibels*) that could be used in conjunction with multi-fingered prostheses endowed with current sensors as those nowadays available like the Touch Bionics i-Limb [28] and RSL Steeper BeBionic hands [29] or research prototypes like the SmartHand [30] or Vanderbilt University Hand [31]. Vibels are composed of N miniature vibration motors having different vibration amplitudes. Therefore for each vibel it is possible to vary both its vibration frequency (varying the strength of the driving current) and to some extent its vibration amplitude (varying the combination and number of active motors). Compared to the state of the art there are two novelties in this present system: first, the architecture allows simultaneous variation of both amplitude and frequency using low-cost components and traditional techniques; secondly the small size allows the integration of the system within a prosthetic socket. Therefore, this system could be easily integrated within a prosthesis and be used for conveying modulated mechanical vibration on sensory target sites in the stump, in a very flexible way. Every time the finger of the prosthesis touches an object a tactile stimulation would be instantaneously delivered to the stump, thereby tricking the brain into experiencing the sensation of touch from the artificial finger [32].

The final goal of this research is to provide acceptable and physiological feedback and thus to make the prosthesis be felt like a part of amputees' body scheme. This paper presents the architecture and the basic principles of operation of a new concept. The implementation on a working prototype and results from preliminary experimentation on healthy subjects are also presented.

II. Architecture and Mathematical Description

A vibration element (vibel) is composed of N vibration motors free to vibrate that can be connected together in a planar or coaxial fashion, as depicted in Fig. 1A and B, respectively (3 vibrators are illustrated in the pictures, as the number used in this work). The N vibration motors might have identical properties or be different amongst them.

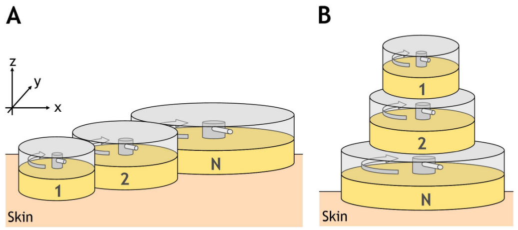

Fig. 1 Vibel architecture: N vibration motors (3 in the picture) are mounted in a planar (panel A) or coaxial (panel B) fashion. In this picture, in both cases, the vibration force is applied tangentially to the skin (shear force).

Each vibration motor consists of a miniaturized electrical motor of mass M with a rotating eccentric mass m mounted in a shaft and bearings; in mechanics this is a classic example of a device generating forced vibrations. The rotation of the unbalanced shaft, induced by the flow of electrical current causes a forced vibration of the motor itself, having a frequency inverse to the rotor revolution period, and vibration force amplitude A proportional to the weight of the eccentric mass m, to its eccentricity e, and to the square of the angular speed, ω. Considering only the displacement on the x-axis in Fig. 1, each of N vibration motors oscillates in a sinusoidal waveform, which force can be generically modelled as a single frequency tone:

$$F=\left(m\cdot e\cdot\omega^2\right)\sin(\omega t)=A\sin(\omega t) \tag{1}$$

and -neglecting second order interferences- but including possible phase shifts ϕ_i, the vibel output force is the sum:

$$Fvibel=\sum\nolimits_{i=1}^{N}A_i\sin(\omega_i t+\phi_i) \tag{2}$$

with i referring to each of N vibration motors. Therefore, generically speaking the output is a complex N-tones oscillation easily described through frequency (spectral) representation but with unintuitive temporal interpretation. A simpler two-tone combination problem (i.e. N=2, A_1=A_2=1) is more easily illustrated in a time framework, under certain restricted hypotheses, as follows. With constant amplitude A, through the prosthaphaeresis formulas and neglecting phase shifts we have:

$$\sin(\omega_1 t)+\sin(\omega_2 t)=2\cos\left(\frac{\omega_1-\omega_2}{2}t\right)\sin\left(\frac{\omega_1+\omega_2}{2}t\right)=$$

$$=2\cos(\omega_{LF}t)\sin(\omega_M t) \tag{3}$$

with ω_{LF} (lower frequency tone) half the difference between ω_1 and ω_2, and ω_M (higher frequency tone) the mean angular speed value. In other words we obtain interference between the two tones, resulting in a modulation of the mean frequency sinusoid ω_M at the lower frequency rate ω_{LF} (cf. Fig. 2A). If the tones have slightly different frequencies this interference is often called *beat*. Another result is achieved when both the amplitude A, and angular speed ω are fixed; in such case, again through prosthaphaeresis formulas, we have:

$$\sin(\omega t)+\sin(\omega t+\phi)=2\cos\left(\phi/2\right)\sin\left(\omega t+\phi/2\right) \tag{4}$$

i.e. a delayed (ϕ/2) and attenuated [by a $2\cos(\phi/2)$ factor] waveform compared to the original one. If the phase shift $\phi\approx 0$, the waveform has about double amplitude with respect to the original one (i.e. *constructive interference*), whereas if the argument of the cosine is ±π/2 the output becomes zero (i.e. *destructive interference*).

In the case of combination of three mechanical vibrations (e.g. as in this work), the time-domain mathematical description becomes significantly heavy, and in simple words all frequency combinations modulate the output waveform; two examples are graphically presented in Fig. 2B and C. However, if the amplitude A, and angular speed ω are fixed, the combination of vibrations generates a waveform with same angular speed of original ones and with a force amplitude that can be the algebraic sum of the three amplitudes (if the phase shifts are zero) or an attenuated version of it as for Equation (4). With reference to Fig. 1 the vibel rotating force vector lies on the x-y plane, hence providing a shear force on the skin; however, the vibel could be placed orthogonally in order to apply normal vibration force.

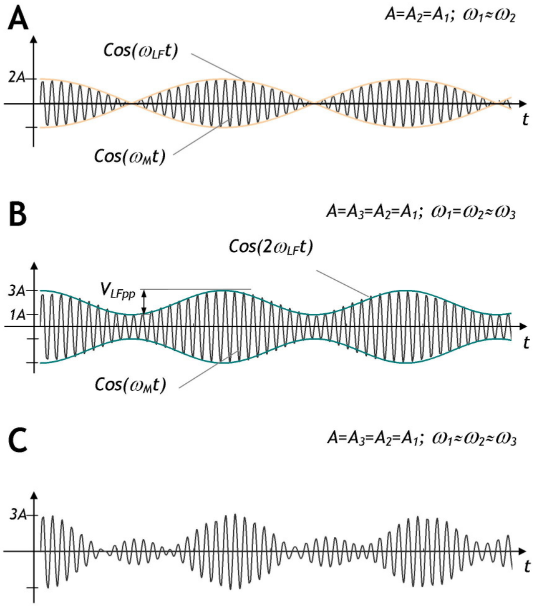

Fig. 2 Time-domain representations of the sum of two or three sinusoidal vibrations. A) *Beat*: the sum of two vibrations with equal amplitude A and different frequencies ω_1 and ω_2, gives a sinusoid that angular speed ω_M is equal to the average between ω_1 and ω_2, and an amplitude of $2A$ which is modulated by a sine having angular speed ω_{LF} half the difference between ω_1 and ω_2 (depicted by the waveform envelope). B) Sum of three vibrations, two of which identical and the third having equal amplitude but different angular speed. This time the envelope modulates at twice ω_{LF}. C) Sum of three vibrations having equal amplitude but different frequencies: all combinations of frequencies contribute to the output waveform.

Preliminary experimental tests with the device confirmed that if the speed of two vibrating motors is significantly different, the output follows Equation (3) and beat interferences are produced (e.g. the waveform shown in Fig. 2A). Other tests revealed that a **coaxial mounting** permits the vibrations from each motor to **constructively combine**, avoiding thus the interference caused by slight differences in frequencies. This effect turns out because the rotor of the miniature motor is magnetically polarized, and hence can magnetically couple with other rotors sufficiently close, like those on the same rotational axis and rotating at similar speed. Therefore, even if the rotational speeds would naturally be slightly different, the rotors are always magnetically synchronized, and the resulting vibration has a greater amplitude (compared to the single motor amplitude) and a unique frequency component. This is an important feature of the coaxially mounted vibel, as it allows to selectively generate beat or constructive interferences. A constructive interference would otherwise be impossible to achieve without accurate control of the speed and rotation of each motor and without synchronization of all rotors. As a proof of evidence (not shown) when placed on the plane (as depicted in Fig. 1A) even if driven with the same current, the vibration motors always generate a beat interference. This is because there is no way for the rotors to synchronize.

III. MATERIALS

In the present work each vibel was composed of three identical miniature motors (Precision Microdrives, UK) (12 mm diameter, 3.4 mm height, 1.7 g mass), each independently providing identical force amplitude A. We considered that three was a good trade-off between dimensions/power demand and different kinds of information that could be conveyed. Motors were mounted coaxially (Fig. 1B) allowing vibels to selectively generate constructive vibrations or beat interferences (like those in Fig. 2). Three vibels were used in this study.

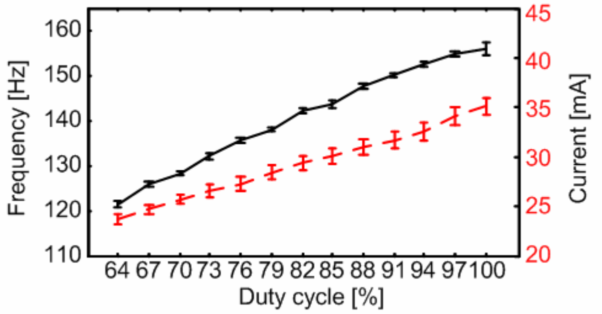

Fig. 3 Vibration frequency and current (striped curve; right Y axis) versus pulse width modulation duty cycle delivered by the microcontroller. Error bars denote the standard error (on 9 measurements).

A microcontroller based electronic board was developed to drive the vibels. Each miniature motor was controlled through a MOSFET used as a switch and driven using a PWM (Pulse Width Modulation) signal generated by the microcontroller. In this way the driving current flowing through the miniature motor, and hence its vibration frequency and the produced force amplitude (cf. Eq. 1), were proportional to the PWM duty cycle. In turn the microcontroller was controlled by a Personal Computer through a RS232 serial interface, communicating with it through a communication protocol developed. This allowed us to select for each vibel: (i) the frequency, (ii) the timbre (the content of harmonics in the spectral representation), (iii) the duration of the stimulation and the spatial location (i.e. which of the vibels or pattern of vibels was active). The delay time from the RS232 command transmission to the effective beginning of the vibration was measured and was lower than 2 ms.

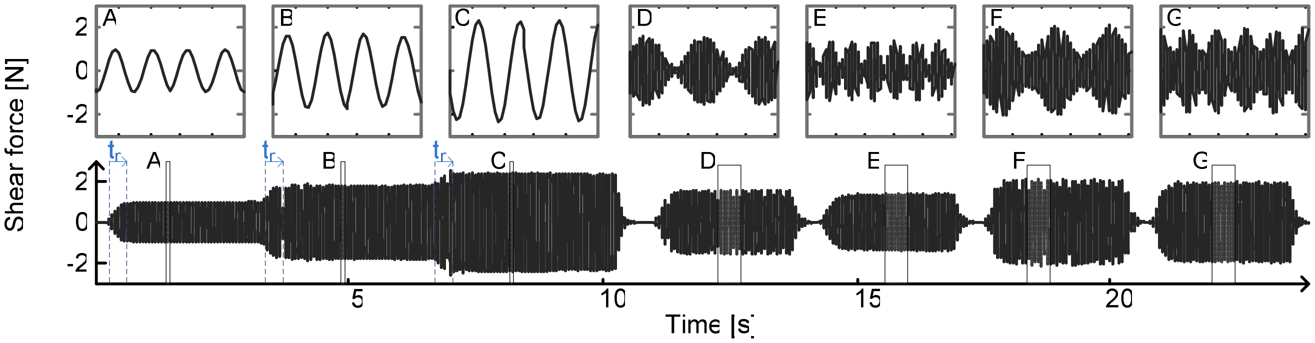

Fig. 4 Representative stimulation sequence allowed by one vibel. A) 1 motor @ 156 Hz; B) constructive interference: 2 motors @ 156 Hz; C) constructive interference: 3 motors @ 156 Hz; D) beat: 1 motor @ 156 Hz and 1 motor @ 140 Hz; E) beat: 1 motor @ 156 Hz and 1 motor @ 122 Hz; F) beat: 2 motors @ 150 Hz and 1 motor @ 140 Hz; F) beat: 2 motors @ 156 Hz and 1 motor @ 122 Hz. The rise time (t_r) ranges between 350-450 ms depending on the final stimulation value.

A software application for testing the vibels with human subjects was developed using LabVIEW (National Instruments). The software allowed us during the experiments to: (i) randomly select the stimulation to be delivered; (ii) provide visual feedback to the subjects during the training, and (iii) log the responses of the subjects in the different experiments.

Fig. 3 shows the required current and the generated output frequency characteristics versus the driving PWM duty cycle for the vibrators used, measured by means of a 6 axis load cell (nano43, ATI, NC, USA). The frequencies allowed by the vibrators chosen, ranged within 112 ± 1 Hz (mean ± standard error, based on 9 measures) and 156 ± 2 Hz, hence in this study three nominal frequencies were selected and tested: 122, 140 and 156 Hz. The representative time plot in Fig. 4 demonstrates a sequence of different stimulations allowed by the developed vibel: in the first ~10 seconds the constructive interferences among the vibrations of 1, 2 and 3 motors are shown (as governed by Equation 4); thereafter the beat interference is highlighted (as in Fig. 2A-B and Equation 3). The plot also shows the rise time, t_r, i.e. the time required to the vibration to reach its steady state; this ranged within 350-450 ms and depended on the final value.

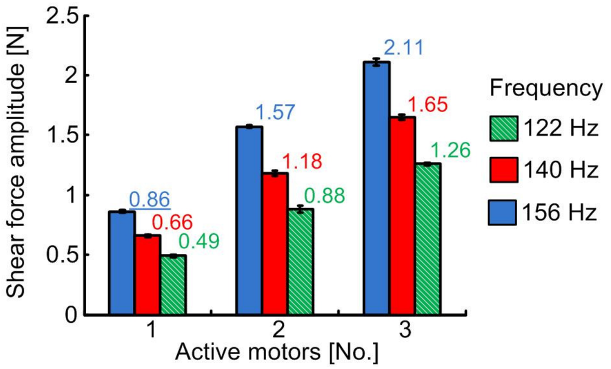

Fig. 5 Shear force produced by constructive interferences of a number of motors active in the vibel for different frequencies. The error bars denote the standard error (9 measurements using a 6-axial load cell).

The graph in Fig. 5 shows the measured amplitude (mean ± standard error based on 9 measures, using the 6 axis load cell) of the shear force produced by 1, 2 or 3 motors inside the vibel at the 3 selected frequencies (constructive interferences). The amplitudes produced varying the number of vibrators, denote that the interferences among vibrations are not perfectly constructive (the amplitude of 2 or 3 vibrators is not twice or three times the amplitude of a single vibrator), i.e. that phase shifts (see Equation 4) are present among the oscillations (around 13 degrees between 1 and 2 vibrators, and around 23 degrees between 2 and 3).

IV. METHODS

The capability of able bodied subjects to discriminate different kinds of stimulation that our vibels were able to produce was assessed through experimental sessions similar to [15]. Although the vibel allowed delivery of beat interferences, this work focused on stimulations only due to constructive interferences. Vibels were placed on the volar aspect of the forearm, kept in place by an elastic sleeve, maintaining the distance from the proximal end of the radius for all subjects (Fig. 6). During the experiments subjects sat with their dorsal forearm placed downward on the bench top and white noise was delivered by headphones in order to mask any auditory stimulation associated with the mechanical sound of the motors. Five different experiments were performed as described in detail in the following paragraphs.

A. *3-Amplitudes Discrimination*

The aim of this experiment (*experiment 3A*) was to evaluate the subjects' ability to discriminate among three different vibration amplitudes allowed by the vibel when one, two or the three miniature motors therein were simultaneously activated to produce a constructive interference (i.e. a single frequency tone). Nine subjects participated, and one vibel was placed on their forearm (Fig. 6, position no. 2). The experiment followed the so called *2 alternatives forced choice* paradigm, described in [33], that consisted of a number of trials in which two vibrations with fixed (and single) frequency but different amplitude were delivered in a sequence. The maximum (156 Hz) and minimum (122 Hz) allowed frequencies were evaluated. The duration of each stimulation and the pause between them was 2.6 s and 1.3 s, respectively (similar to [23]). Both orders of presentation (higher or lower amplitude first) were used and all combinations of amplitudes (1 vs. 2 motors, 1 vs. 3 and 2 vs. 3) were tested 10 times for each frequency. Hence, for each subject 60 trials were randomized among the conditions. After each trial the subject was asked to declare which vibration (first or second) was perceived with higher intensity. For the purpose of data analysis, subjects were considered capable of successfully discriminating among amplitudes if the percentage of correct response was above 75% (i.e. a 75% discrimination threshold).

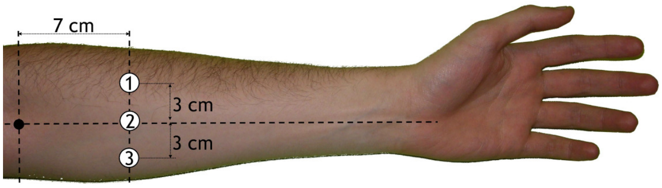

Fig. 6 Vibels placement on the volar aspect of the forearm during the experimental evaluation. Vibel number 2 was used for the amplitude discrimination (*experiment 3A*) and frequency-amplitude discrimination tests. (*experiment A* and *experiment F*). The three vibels were used in the sites (*experiment S*) and patterns (*experiment P*) discrimination tests.

B. *Discrimination of Frequency-Amplitude Combinations*

In the eccentric motor both frequency and amplitude of the vibration are proportionally dependent on the driving current (cf. Introduction and Eq. 1). This means that it is not possible to select the frequency independently from the amplitude or vice-versa. To investigate the effects on human perception of this physical property (i.e. intrinsic of the vibration motor), two identical experimental protocols were carried out: we investigated the subjects' ability to perceive (i) purely amplitude differences (*experiment A*) or (ii) purely frequency differences (*experiment F*) when stimulations having simultaneous variations in frequency and amplitude were delivered in a sequence. Nine subjects participated in both the experiments, described as follows.

One vibel was placed on the forearm (Fig. 6, position no. 2) and in each trial two vibrations were delivered; these consisted of a *standard* stimulation, the same for all trials, and a second, *comparison* stimulation, having different frequency and amplitude. The standard stimulus had the maximum frequency (156 Hz) and lowest amplitude allowed by the vibel (only one motor was activated). The comparison stimulus could vary among six different ones: it could be produced by one, two or three motors (in the vibel) at the frequencies of 140 Hz or 122 Hz. Hence, six different comparison force amplitudes (of interest for *experiment A*) were tested. The two comparison stimulus frequencies (of interest for *experiment F*) were selected in order to test different conditions: an easier one (difference between standard and comparison greater than 30 Hz) and a more difficult to discriminate (difference < 30 Hz). Each of the six comparison stimuli was tested 10 times and during the experiment the 60 trials were randomized. Both orders of presentation (standard vs. comparison and comparison vs. standard stimuli) were used. The length of and pause between the two stimulations were those used in the amplitude discrimination experiment. After the trial, based on which experiment was being performed, the subject determined which stimulation was perceived with higher frequency (in *experiment F*) or higher amplitude (in *experiment A*). Subjects were also instructed to ignore, as far as it was possible, the difference between standard and comparison stimuli of the other parameter, hence giving an answer based only on the frequency (*experiment F*) or amplitude perception (*experiment A*). A 75% discrimination threshold as in the previous test was used.

C. *Sites and Patterns Discrimination*

Finally, two different experiments aimed to evaluate spatial discrimination (among three different sites, *experiment S*) and pattern discrimination (i.e. combinations of stimuli with two different intensities on three different sites to simulate six different grips by a robot prosthesis, *experiment P*) were performed consecutively. Ten able-bodied subjects were involved in both the experiments (same subjects in both experiments). Three vibels were placed on the forearm along the ulnar-radial direction (cf. Fig. 6), with 3 cm spacing. This distance was chosen based on preliminary two point discrimination tests (85% discrimination percentage) [34].

Both *experiment S* and *experiment P* were divided into three sessions: a learning session with visual feedback, a reinforced learning session without vision, and a validation session also without vision. With regards to *experiment S,* during the learning session, the participant watched a computer screen showing which vibel was stimulating his/her forearm synchronously. This session consisted of 48 stimulations at 156 Hz: among these, on each site, 8 stimulations were provided by 1 motor, and 8 by 3 motors (constructive interference); each stimulation was active for 2.6 seconds. In the reinforced learning session, the participant was blindfolded; after the presentation of each stimulus, the participant verbally indicated the stimulation site and the test supervisor stated the correct answer, hence reinforcing the learning. As in the previous session the number of stimulations was 48. The last session consisted of 42 stimulations and was used to validate the results of the learning and reinforced learning sessions. This setup was based on previous similar studies [12], [15].

TABLE I
COMBINATIONS OF SITE-INTENSITY STIMULATIONS MIMICKING SIX GRASPS

ID	Site	No. of active motors (156 Hz)	Mimicked grasp
1	1	1	Light-force one-finger grasp (e.g. lateral grip)
2	1	3	High-force one-finger grasp (lateral)
3	1,2	1	Light-force two-finger grasp (e.g. pinch)
4	1,2	3	High-force two-finger grasp (pinch)
5	1,2,3	1	Light-force 3-finger grasp (e.g. cylindrical)
6	1,2,3	3	High-force 3-finger grasp (cylindrical)

In the pattern discrimination experiment, *experiment P,* the setup was similar: the participant had to learn six different stimulation patterns simulating the effects of six grasps by an artificial prosthesis (cf. Table I). 42 stimulations (7 for each pattern) were used for the learning, reinforced learning and validation sessions.

V. RESULTS

A. 3-Amplitudes Discrimination

Fig. 7 shows the mean percentage of discrimination for all subjects, for both the frequencies and all combinations of stimuli (1 vs. 2 motors, 2 vs. 3 motors and 1 vs. 3 motors). The mean discrimination percentage was above 75% discrimination threshold. The amplitude discrimination percentage for 1 vs. 2 motors was between 75 and 80 %, in the other two cases it was greater than 90 %. The 3-way ANOVA showed that the difference in the responses across the two frequencies and the subjects, was not statistically significant; statistical differences were found instead among the combinations of stimuli ($p < 0.001$).

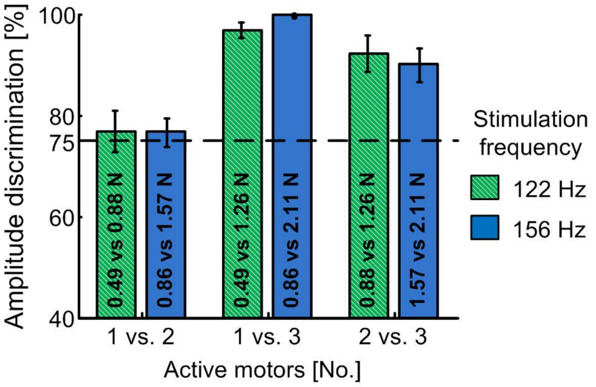

Fig. 7 Mean discrimination percentages (± standard error) for all subjects in *experiment 3A*. The different force amplitudes in the different cases are superimposed on the bars. Striped bars refer to 122 Hz and plain bars to 156 Hz stimulations.

B. Discrimination of Frequency-Amplitude Combinations

Table II presents the correct answer percentages for each subject in *experiment F.* In eight subjects (out of nine) the greatest correct discrimination (> 70%) between standard and comparison stimuli was achieved when 1 single vibrator provided the comparison stimulus, i.e. when the comparison forces were lower than the standard ones. Specifically the differences were equal to 0.20±0.02 N for the 144 Hz and 0.37±0.02 N for the 122 Hz comparison stimulations. The more vibrators being active the worse discrimination being achieved. Only in one subject (i.e. AM) this trend was found to be opposite (i.e. the highest discrimination percentages were achieved when 3 motors provided the comparison stimulation): this contradictory result is actually in accordance with similar psychophysical studies, such as Morley and Rowe [23]. Indeed, they showed that some individuals perceive increased amplitudes as increased frequencies, whereas other individuals perceive them in the opposite way (increased amplitudes felt as decreased frequencies). The percentage of correct discrimination for the 122 Hz comparison stimulation delivered by 2 motors was also above 70% for all subjects; in this case the force amplitudes of the comparison and standard stimulations were similar (mean difference: 0.02±0.04 N).

The mean discrimination percentage, considering all subjects excepting AM (who showed a perception trend opposite to the others, hence did not belong to the same group), resulted above the 75% threshold only when the amplitude of comparison stimulation was lower than or close to the standard one. A close to 25% threshold (i.e. 30 ± 9 %), meaning that almost 75% of the trials were judged opposite, was achieved when the comparison amplitude was greatly higher (almost double) than in the standard stimulation (i.e. comparison stimulation by 3 motors @ 140 Hz). The 3-way ANOVA demonstrated no significant difference across the discrimination percentages of subjects. The difference was significant across comparison frequencies ($p < 0.001$) and vibration amplitudes ($p < 0.001$).

TABLE II

CORRECT RESPONSES (%) IN FREQUENCY DISCRIMINATION TRIALS

Subj.	**Comparison frequency: 122 Hz**			**Comparison frequency: 140 Hz**		
	1 motor	*2 motors*	*3 motors*	*1 motor*	*2 motors*	*3 motors*
PM	100	100	30	80	40	10
GB	90	80	40	100	70	20
MC	100	80	30	100	60	20
MCf	100	90	50	70	10	50
MD	80	70	40	80	30	60
LV	90	70	40	90	20	0
MF	80	70	50	90	10	10
IM	100	100	80	80	50	70
AM	20	70	70	20	50	100
Mean ± S.E. *	92 ± 3	82 ± 5	45 ± 6	86 ± 4	36 ± 8	30 ± 9

*mean value and standard error excluding AM.

The outcomes for *experiment A* are reported in Fig. 8; the graph shows the mean correct response percentages, ordered on the X axis, based on the difference between the standard and comparison stimulations. The horizontal dashed axes denote the 25% and 75% thresholds, whereas the vertical one marks the amplitude of the standard stimulation. Given the nature of the 2 alternatives forced choice, when the standard and comparison stimulations coincide, the response on "which stimulation was perceived with a higher amplitude" yields to a average answer of 50% in favor of the first (e.g. the standard) and 50% in favor of the second stimulation (e.g. the comparison). In other words, when the two stimulations are very similar, statistically, 50% of the times one and the other are perceived to have higher amplitude.

The graph in Fig. 8 demonstrates successful discrimination (above the 75% threshold) when the comparison stimulation was delivered by one or three motors in the vibel, regardless of the frequency (both 122 Hz and 140 Hz). Interestingly, a close to the 25% threshold (i.e. 27 ± 4 %), meaning that 75% of trials were felt in the opposite way, was achieved when the comparison stimulation was provided by 2 motors vibrating @ 122 Hz. In addition, a close to the 50% value (i.e. 46 ± 7 %) was achieved when the comparison stimulation was delivered by 2 motors vibrating @ 140 Hz; this means that standard (0.86 N @ 156 Hz) and comparison (1.18 N @ 140 Hz) were statistically considered as the same stimulation.

The 3-way ANOVA showed that the difference in the responses among the subjects was not significant. The difference was significant between the comparison frequencies ($p < 0.01$) and among the vibration amplitudes ($p < 0.001$).

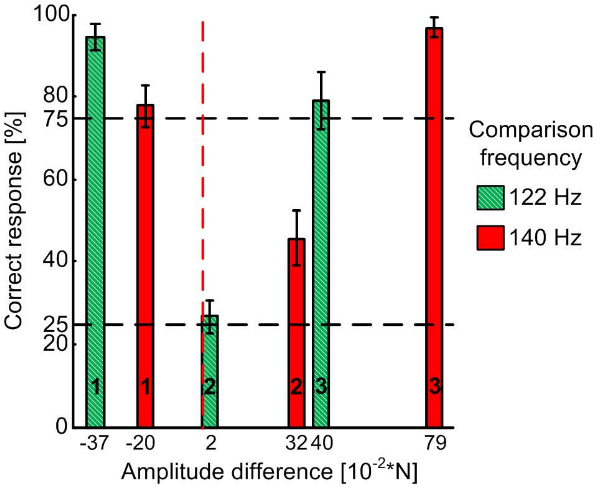

Fig. 8 Correct response percentages (± standard error) for *experiment A*. Striped bars refer to 122 Hz and plain bars to 140 Hz stimulations. The vertical striped line marks the standard stimulation vibration force amplitude. Figures superimposed on the bars indicate the number of active motors used to provide such comparison stimulation.

C. *Sites and Patterns Discrimination*

Results relative to *experiment S* are presented in Fig. 9 (left side graphs). The upper and lower confusion matrices show the reinforced learning and the evaluation session outcomes, respectively. The X axes denote the stimulation site (with reference to numbers in Fig. 6); the Y axes denote which site was felt being stimulated (the responses). The intensity of the gray-scale represents the frequency of correct answers related to the stimulation. Both during the learning and evaluation phases adjacent stimulation sites were occasionally mixed up (1 vs. 2, 2 vs. 3). However the percentage of correct responses of the evaluation session (i.e. when the subject is supposed to be better trained) (93 ± 2 %) was higher than in the reinforced learning session (90 ± 2 %). The 2-way ANOVA demonstrated significant difference between reinforced learning and evaluation data ($p < 0.01$). Conversely, the difference in recognition percentage between different stimulation amplitudes was not significant.

The results of the pattern discrimination tests (*experiment P*) are presented in Fig. 9 (right side graphs). Numbers on the horizontal and vertical axes denote the stimulations as identified in Table I. The average correct response percentage during reinforced learning session was equal to 66 ± 4 %, and significantly different ($p < 0.001$) from the evaluation session performance, equal to 78 ± 3 %. Interestingly, during the reinforced learning session the most confused patterns were no. 2, 4, and 5; whereas during the evaluation session they were *adjacent* patterns, i.e. patterns: 1 vs. 2, 2 vs. 3, 4 vs. 5 and 5 vs. 6, and in addition, pattern 2 vs. 4 (showing the lowest accuracy).

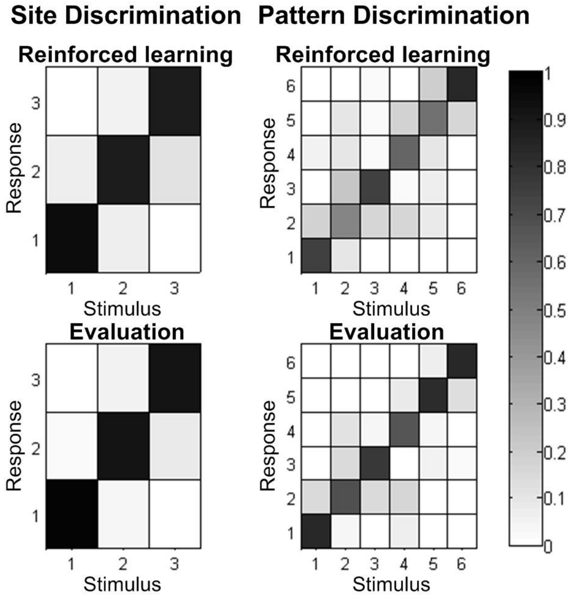

Fig. 9 Mean discrimination ratios for all subjects in *experiment S* (left side graphs) and *experiment P* (right side graphs).

VI. DISCUSSION

This work introduces a new, simple, non-invasive sensory substitution system composed of multiple vibrating elements physically connected, which exploits the constructive or beat interferences that result when two or more vibrations are combined. This work focused on hand prosthetics and on the application of the vibels onto the glabrous skin of the forearm/stump. Nevertheless the vibel architecture could be used for other applications (or other body sites) in which multi-dimensional information needs to be conveyed: e.g. as a means for providing feedback events or cues for rehabilitation systems, navigation guidance systems for blind, in video-games controllers, etc. The clear advantages of the vibel architecture are: (i) extremely low cost (< 1€), (ii) miniaturized dimensions and (iii) low power demand (e.g., a 3.3 V, 0.32 Ah battery would suffice for a one hour continuous vibration of three vibels at maximum frequency-force level). Even though a vibel can provide a two dimensional information, and makes it possible to selectively produce beat (cf. Fig. 4D-G) or constructive interferences (cf. Fig. 4A-C) with no need for other components other than the vibrators and a simple controller.

The main drawback of vibrotactile systems, and hence of this device, is the rise time, i.e. the time needed to reach the steady state of the desired stimulation. Since vibration is produced by a moving mass, it is necessary to provide enough energy to counteract its inertia: the heavier the mass, the higher are the reaction times (cons) as well as the produced vibration amplitudes (pros). In this specific prototype, to produce 0.86 – 2.86 N shear force amplitudes, the rise time ranged between 350 – 450 ms. Such delays are considerably high with regards to the timing requirements of a prosthesis intended to replace the lost hand (in which afferent stimulation should occur instantaneously with the triggering event), and hence, tradeoffs between reaction times and producible forces are necessary. For the sake of clarity however, it should be noted that the sensation threshold in the forearm is lower by orders of magnitude compared to the forces used in this study and in the range of 10^{-3} – 10^{-2} N [8]; therefore since sensation threshold forces are almost instantaneously reached by the vibels (~ 1 - 10 ms), it is possible to state that our device is able to produce real-time cues, fundamental in a bi-directionally interfaced prosthesis [35]. Anyway, the delays of the present system could not have influenced the experimental outcomes of this work.

Although the vibel architecture allows delivery of beat interferences, this work only investigated the ability by healthy volunteers to perceive vibrations with different amplitudes and frequencies due to constructive interference. The perception of beat will be investigated in future research.

Experiment 3A showed that healthy subjects were able to discriminate (success > 75 %) among the three amplitudes generated by constructive interferences (as shown e.g. in Fig. 4A-C). In addition since no statistically significant differences in discrimination percentages across the two frequencies were found, it can be concluded that within the frequency range 122-156 Hz the capability of subjects to discriminate different shear forces (at constant frequency) does not change. It should be noted that the range of operation allowed by the vibels in this study (112 -156 Hz) depended on the choice of the miniature vibrators. A different selection of miniature vibrators will yield to different performance (and power demands).

Experiment F and *experiment A* confirmed previous studies: perceived frequency (or amplitude) is affected by amplitude (or frequency) changes [23]. Within the ranges here investigated, if frequency and amplitude of standard and comparison vibrations vary concurrently, high discrimination ability is achieved only if this variation is coherent, i.e. both increase or decrease. In other words, a decreased frequency (or amplitude) is properly perceived only if the amplitude (or frequency) has not increased. In the latter case, depending on the magnitude of the variations, subjects can either be confused and unable to clearly judge (percentages between 25-75 %), or be statistically wrong (percentages close to 25 %), i.e. perceive the variation of the other parameter. The effects of concurrent frequency/amplitude variations (caused by the physical nature of the motors with eccentric mass) however can also be positively exploited: with regards to the frequency discrimination, previous studies revealed that in a range included between 100 and 200 Hz a human subject was able to correctly discriminate (about 50% discrimination accuracy through psychometric function curves analysis) a difference of at least 30 Hz [21]. In this study, when the actions of amplitude and frequency were coherent (both decreasing/increasing) the standard (156 Hz) and the (140 Hz) comparison stimulations, differing only 16 Hz, were correctly discriminated. In other words, the concurrent and coherent variation of the two components virtually improves the discrimination ability, and hence, a greater number of differently perceived stimulations can be found.

Experiment S showed that subjects were able to discriminate among three vibels placed on the skin of their forearm volar aspect. The discrimination was independent from the number of motors activated in the vibels and the high (> 90 %) recognition percentages achieved both during learning and training phases, demonstrated the intuitiveness of the system from the beginning of tests, and the short training procedure required. When misclassified, sites were confused with adjacent ones (1 vs. 2 and 2 vs. 3) both during the reinforced learning and the evaluation phases. Misclassified responses can be explained considering the short distance (3 cm) between nearby sites, close to the two-point discrimination of the forearm [36], [8]. Other studies on the capability of localizing vibrotactile stimuli on the forearm were carried out in the past: Cholewiak et al. [37] developed and studied a 7-elements piezoceramic tactor array along the volar part of the forearm, whereas Oakley et al. [38] investigated a 3x3 matrix of miniaturized motors, applied on the dorsal part of the forearm. Both studies used a 2.5 cm distance between the elements, and both reported a mean localization rate close to 50%, i.e. significantly lower than our results (> 90% for a 3 cm distance). This difference is likely due to the different number of elements used: the larger the number of choices (of sites), the more complex is the discrimination.

The most promising results are those achieved in *experiment P* i.e. the pattern (grasp) discrimination experiment. Although the outcomes demonstrated lower success in both reinforced learning and evaluation phases (i.e. higher perceptual difficulty) compared to *experiment S*, the accuracy in the evaluation phase, reached the significantly high percentage of 78 % among the 6 patterns. In other words, this means that if embedded inside the prosthetic socket and connected to sensors on a dexterous prosthesis, 6 different grasps (associated to a different number of involved fingers and different force levels, as suggested in Table I) could be correctly perceived by the amputee. To demonstrate this hypothesis and to investigate the usefulness of the proposed architecture on amputees, further work is planned.

Multi-fingered prostheses endowed with artificial tactile sensibility are progressively becoming the reality [27]-[31], and sensory feedback systems able to convey spatially distributed afferent information could improve amputees' awareness of the prosthesis and of the environment, as well as voluntary controllability, and hence quality of life. Reaching this goal i.e. *what type* and *how much* information should be provided, is still an open question, impossible to tackle without a practical validation. In the authors' opinion, for enriched environment perception through an artificial sense of touch, at least three types of basic information need to be tailored to the amputee: 1) the contact cue, instantaneously delivered with the tactile triggering event (*when* touch happens); 2) the tactile-map, i.e. roughly *where* the touch is taking place (e.g. which fingers, and/or palmar areas) and 3) the tactile force, i.e. *how* much force is applied on the involved fingers and palmar areas (denoting the tactile-map) in the prosthesis.

ACKNOWLEDGMENT

Authors would like to thank P. Stabile, S. P. Murdoch, and C. Antfolk for their precious help, as well as the volunteers that took part in the experiments.

REFERENCES

[1] M. A. Oskoei, H. Hu, "Myoelectric control systems - a survey," *Biomed. Signal Process. Control*, vol. 2, pp. 275- 294, 2007.

[2] P. J. Kyberd C. Wartenberg, L. Sandsjo, S. Jonsson, D. Gow, J. Frid, C. Almstrom, and L. Sperling, "Survey of upper extremity prosthesis users in Sweden and United Kingdom," *Journal of Prosthetics and Orthotics*, vol. 19, pp.55–62, 2007.

[3] E. Biddiss, D. Beaton, and T. Chau, "Consumer design priorities for upper limb prosthetics," *Disability and Rehabilitation: Assistive Technology*, vol. 2: 6,pp. 346-357, Nov. 2007.

[4] G. Lundborg and B. Rosen, "Sensory substitution in prosthetics," *Hand Clin.,*vol. 3,pp. 481 488, 2001.

[5] D. S. Childress, " Closed-loop control in prosthetic systems: historical perspective," *Annals of Biomedical Engineering*, vol. 8, pp. 293-303, 1980.

[6] G. S. Dhillon and K. W. Horch, "Direct neural sensory feedback and control of a prosthetic arm," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 13, no. 4, pp. 468–472, 2005.

[7] P.M. Rossini, S. Micera, A. Benvenuto, J. Carpaneto, G. Cavallo, L. Citi, C. Cipriani, L. Denaro, V. Denaro, G. Di Pino, F. Ferreri, E. Guglielmelli, K-P. Hoffmann, S. Raspopovic, J. Rigosa, L. Rossini, M. Tombini, and P. Dario, "Double Nerve Intraneural Interface Implant on a Human Amputee for Robotic Hand Control," *Clinical Neurophysiology*, vol. 121, no. 5, pp. 777-783, 2010.

[8] K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita, and W. J. Tompkins "Electrotactile and vibrotactile displays for sensory subtitution system," *IEEE Trans. Biomed. Eng.*, vol. 38, pp. 1-6, Jan. 1991.

[9] B. Geng, K. Yoshida, W. Jensen, "Impacts of selected stimulation patterns on the perception threshold in electrocutaneous stimulation," *Journal of NeuroEngineering and Rehabilitation,* vol. 8, no. 9, 2011.

[10] G. F. Shannon, "A comparison of alternative means of providing sensory feedback on upper limb protheses," *Med. Biol. Eng.* ,vol. 14, pp. 289-294, 1976.

[11] O. Sueda and H. Tamura, "Sensory device for the artificial arm," *In Proc. of the eighth Int. Conf. on Med. And Biol. Eng.*, Chicago, 1969.

[12] C. Cipriani, C. Antfolk, C. Balkenius, B. Rosén, G. Lundborg, M. C. Carrozza and F. Sebelius, "A Novel Concept for a Prosthetic Hand with a bidirectional Interface: A Feasibility Study," *IEEE Trans. Biomedical Engineering*, vol. 56, no. 11, part 2, pp. 2739-2743, 2009.

[13] S. G. Meek, S. C. Jacobsen, and P. P. Goulding, "Extended physiologic taction: design and evaluation of a proportional force feedback system," *J Rehabil Res Dev*, vol. 26, pp. 53-62, Summer 1989.

[14] H. Sears, E. Iversen, S. Archer, J. Linder, K. Hays, "Grip force feedback in an electric hand – preliminary results," *In: Proc. Myoelectric Controls/Powered Prosthetics Symposium*, Fredericton, NB, Canada, pp. 175-178, 2008.

[15] C. Antfolk, C. Balkenius, B. Rosén, G. Lundborg, and F. Sebelius, "SmartHand tactile display: A new concept for providing sensory feedback in hand prostheses," *Scandinavian Journal of Plastic and Reconstructive Surgery and Hand Surgery*, vol. 44, pp. 50-53, 2010.

[16] Q. An, Y. Matsuoka, C. E. Stepp, "Multi-day Training with Vibrotactile Feedback for Virtual Object Manipulation," *In Proc. IEEE RAS/EMBS Intl. Conf. on Rehabilitation Robotics*, ICORR, Zurich, Switzerland, June 29 – July 1, 2011.

[17] K. Kim, J. E. Colgate, J. J. Santos-Munnè, A. Makhin and M.Peshkin, "On the design of miniature haptics devices for upper extremity prosthetics," *IEEE Trans. Mechatronics*, vol. 11, pp. 27-39, 2010.

[18] P. D. Marasco, K Kim, J. E. Colgate, M. A. Peshkin and T.A. Kuiken, "Robotic touch shifts perception of embodiment to a prosthesis in targeted reinnervation amputees," *Brain*, vol. 134, pp. 747-758, 2011.

[19] M. M. Merzenich and T. Harrington, "The sense of flutter- vibration evoked by stimulation of the hairy skin of primates: comparison of human sensory capacity with responses of mechanoreptive afferents innervanting the hairy skin of monkeys," *Exp. Brain Res.*, vol. 9, pp. 236-260, 1969.

[20] E. Francisco, V. Tannan, Z. Zhang, J. Holden, and M. Tommerdahl, "Vibrotactile amplitude discrimination capability parallels magnitude changes in somatosensory cortex and follows Weber's law," *Exp. Brain. Res.*, vol. 191, pp. 49-56, 2008.

[21] D. A. Mahns, M. Perkins, V. Sahai, L. Robinson, and M.J. Rowe, "Vibrotactile frequency discrimination in human hairy skin," *J Neurophysiol*, vol. 95, pp.1442- 1450, 2006.

[22] R. T. Verrillo, "Psychophysics of vibrotactile stimulation," *J. Acoustical Society America*, vol. 77, pp. 225-232, 1985.

[23] J. W. Morley, and M.J. Rowe, "Perceived pitch of vibrotactile stimuli: effects of vibration amplitude, and implications for vibration frequency coding," *J. Physiol*, vol. 431, pp. 403-416, 1990.

[24] A. Chatterjee, P. Chaubey, J. Martin, and N. Thakor, "Testing a prosthetic haptic feedback simulator with an interactive force matvhing task," *J. of Prosthetics and Orthotics*, vol. 20, pp. 27-34, 2008.

[25] C. Cipriani, F. Zaccone, S. Micera, and M.C. Carrozza, "On the shared control of an EMG controlled prosthetic hand: analysis of user-prosthesis interaction," *IEEE Trans. On Rob.*, vol. 24, pp. 170-184, 2008.

[26] J. L. Pons, E. Rocon, R. Ceres, D. Reynaerts, B. Saro, S.levin, and W. Van Moorlenghem, "The MANUS-HAND dextrous robotics upper limb prosthesis, mechanical and manipulation aspects," *Autonomous Robots*, vol. 6, pp. 143-163, 2004.

[27] C. Pylatiuk, A. Kargov, S. Schulz, "Design and Evaluation of a Low-Cost Force Feedback System for Myoelectric Prosthetic Hands," *Journal of Prosthetics and Orthotics*, vol. 18, no. 2, pp. 57-61, 2006.

[28] Touch EMAS Ltd, Edinburgh, U.K. [Online]. Available: http://www.touchbionics.com

[29] RSL Steeper, Rochester, U.K. [Online]. Available: http://www.bebionic.com

[30] C. Cipriani, M. Controzzi, M. C. Carrozza, "The SmartHand Transradial Prosthesis," *Journal of NeuroEngineering and Rehabilitation*, vol. 8, no. 29, 2011.

[31] S. A. Dalley, T. E. Wiste, H. A. Varol, M. Goldfarb, "A Multigrasp Hand Prosthesis for Transradial Amputees", *In Proc. of Intl. Conf. of the Engineering in Medicine and Biology Society, EMBC*, Buenos Aires, Argentina, Aug. 31 – Sept. 4, 2010.

[32] B. Rosén, H. Ehrsson, C. Antfolk, C. Cipriani, F. Sebelius, G. Lundborg, "Referral of sensation to an advanced humanoid robotic hand prosthesis," *Scandinavian Journal of Plastic and Reconstructive Surgery and Hand Surgery*, vol. 43, pp. 260-266, 2009.

[33] D. McNicol, " A primer of signal detection theory," Allen and Unwin, Sydney, 1972.

[34] M. D'Alonzo, C. Cipriani, M. C. Carrozza, "Vibrotactile Sensory Substitution in Multi-fingered Hand Prostheses: Evaluation Studies," *In Proc. IEEE RAS/EMBS Intl. Conf. on Rehabilitation Robotics*, ICORR, Zurich, Switzerland, June 29 – July 1, 2011.

[35] R. S. Johansson, B. B. Edin, "Predictive feedforward sensory control during grasping and manipulation in man," *Biomed Res*, vol. 14, pp. 95-106, 1993.

[36] S. Weinstein, "Intensive and extensive aspects of tactile sensitivity as a function of body part, sex, and laterality," in The Skin Senses, D.R.Kenshalo, Ed. Springfield, IL: Charles C. Thomas, pp. 195–218, 1969.

[37] R. W. Cholewiak and A. A. Collins, “Vibrotactile location on the arm: effect of place, space and age,” *Percept. & Psycophys.*, vol. 65, pp. 1058-1077, 2003.

[38] I. Oakley, Y. Kim, J. Lee and J. Ryu, “Determining the feasibility of forearm mounted vibrotactile displays,” *Int. Proc. Of Haptics Symp.*, 2006.