NONCONTACT MMG SENSOR BASED ON THE OPTICAL FEEDBACK EFFECT IN A LASER DIODE

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ABSTRACT

An optical interferometric MMG sensor is presented. It uses the feedback effect in a laser diode to measure the vibration generated by the MMG at the skin surface. Its key features are noncontact measurement, high sensitivity, and extended bandwidth toward low frequencies (≈1 Hz). Furthermore, as it is able to recover with a good accuracy the shape of vibrations ranging from 1 μm peak-to-peak, the MMG is expressed in the physiological unit (micrometer) rather than the sensor dependent unit. The MMG measurement conditions are reviewed through a clinical protocol, including the perturbation causes. This probe is compared with a classical microphone based sensor, and thereby the influence of the coupling between the muscle and a contact sensor is demonstrated. © *1998 Society of Photo-Optical Instrumentation Engineers.* [S1083-3668(98)00803-X]

Keywords MMG; muscle vibration; MMG sensor; optical feedback; laser diode.

1 INTRODUCTION

Muscle vibration, whose measurement is generally called MMG, is a mechanical phenomenon generated by the muscle fiber activity. It is a compound signal deriving from the summation of the active muscle fiber twitches.[1] Its time and frequency characteristics depend on the muscle structure,[2] state, and the contraction process[3] through the type and number of recruited motor units, and their firing rate. Since the work of Oster and Jaffe,[4] who initiated the modern investigations of the MMG, there is a growing interest in this phenomenon. It is principally motivated by the present need for noninvasive muscle investigation techniques, which could complete the information provided by the EMG. MMG has been used recently, for example, for muscle fiber typing in sports medicine,[5] and to investigate the effect on the muscle of neuromuscular diseases[6] or age.[7] It has also been shown that it is a good index for force[8] and fatigue[9] for muscles whose force cannot be measured directly, such as the diaphragm.

If the properties of the MMG have been extensively studied, few authors have focused on the measurement conditions, which greatly influence the reliability and the repeatability of the results. A MMG sensor must be able to measure a vibration in the micrometer range, with frequencies from 1 to about 100 Hz.[10] Various transducers have been presented, such as accelerometers,[11,12] piezoelectric contact sensors,[13,14] or air-coupled microphones.[8,9] Except for accelerometers, the MMG is generally not given in physiological units but in sensor dependent units, which limits comparisons between studies. Furthermore, as Orizio has shown, the shape of the measured MMG depends on the sensor.[3] It is influenced by the transducer properties and the mechanical coupling with the muscle. Recently, Courteville et al.[10] described a microphone based MMG sensor calibrated in frequency and amplitude. It uses an acoustic impedance adaptation to reach a high sensitivity while minimizing the constraint on the muscle and the coupling effects. This sensor demonstrated its efficiency as a clinical tool but, like all other existing MMG sensors, it needs to be in contact with the skin.

This article present an optical interferometric MMG probe based on the feedback effect in a laser diode. Its key features are noncontact measurement, high sensitivity, and extended bandwidth toward low frequencies. A comparison with a more classical sensor[10] will also be achieved to investigate, among other things, the effect of the contact between sensor and muscle. Finally, we will review the noise sources that can perturb optical MMG measurement.

2 OPTICAL SENSOR DESCRIPTION

The purpose of the optical sensor is the MMG measurement without contact. It senses the vibration

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generated by the muscle and transmitted to the skin surface. It is a signal in the micrometer range whose frequencies are between 1 and 100 Hz. Optics provide a large variety of interferometric methods to sense such a vibration.[15] They have proven their efficiency for measurements on highly reflecting targets (mirrors). In our case, however, the skin is very absorbing and a scattering medium for which many of these methods are not applicable. Therefore, we have opted for a particular interferometric technique based on the optical feedback effect in a laser diode[16–18] that works as well on highly as on poorly reflecting targets. It derives from the method described by Gharbi et al. in Ref. 19, so the theoretical background will only be briefly reported here.

The light emitted by the laser diode is backscattered on the skin. A fraction re-enter the laser diode cavity. It is well known that lasers, and particularly laser diodes, are extremely sensitive to optical feedback: a reinjected power whose order of magnitude is 10^{-4}–10^{-5} of the emitted power is enough to perturb the emission. The target, located at a distance L_{ext} from the laser diode, forms an external cavity coupled to the laser diode cavity, thus modifying the amplitude and phase conditions of the laser. For low feedback levels, it results in a small variation of the output optical power which can be expressed as[20]

$$\Delta P = \Delta P_{\max} \cos(\omega_c \tau_{\text{ext}}), \tag{1}$$

$$\tau_{\text{ext}} = \frac{L_{\text{ext}}}{c}, \tag{2}$$

where is the external cavity round trip time, c is the light velocity, and ω_c is the laser operating angular frequency for the compound cavity. $\Delta P_{\max}$ is the power variation amplitude, which depends, among other things, on the feedback level. A target vibration results, through Eq. (1) and Eq. (2), in a phase modulation of the optical power variation ΔP. There are many methods to compute the velocity or the displacement from ΔP.[17,18] We have opted for the one described in Ref. 19 because of its simplicity and robustness. Accordingly, the injection current of the laser diode is modulated by a symmetric triangular signal of period T_m. It causes the optical frequency to vary linearly along each modulation ramp. When optical feedback is present, ΔP becomes a periodic signal whose fundamental beat frequency is F_+ on the upward modulation ramp and F_- on the downward ramp. As it has been demonstrated in Ref. 19, the mean target velocity $\langle V \rangle_{T_m}$ over a modulation period T_m can be then expressed as

$$\langle V \rangle_{r_m} = \frac{\lambda_0}{4} [\langle F_- \rangle - \langle F_+ \rangle], \tag{3}$$

where λ_0 is the mean optical wavelength. The displacement is derived by integration of the velocity. Notice that $1/T_m$ also becomes the vibration sampling frequency.

The experimental setup is shown in Figure 1. A Sharp LT026MDO laser diode (785 nm, 3 mW) is used. The beam is collimated and then focused on the skin by a 200 mm focal length lens. The laser bias current is set at 47 mA, and its temperature regulated at 20°C±0.05°C by a thermoelectric cooler. This operating point is chosen to avoid mode hopping. The injection current is modulated by a 2.5 mA peak-to-peak symmetric 500 Hz triangular signal. The optical power P (Figure 2, curve A) is measured by the photodiode incorporated into the laser package. The amplitude of the power variation ΔP due to feedback is only 2% of the overall power modulation amplitude. So, ΔP (Figure 2, curve B) is extracted from P by subtracting the high-amplitude triangular modulation from the optical power. It is then sampled at 500 kHz and fed into a computer.

3 Experimental Setup

3.1 Clinical Setup

The purpose of the clinical setup is to record MMG and force simultaneously during an isometric effort in well defined muscle work conditions. It is based on the protocol described in Ref. 21. The subject adjusts its effort to a preprogrammed consign value. He grasps with his right hand a 4 cm diam cylinder with a strength sensor inside.[21] It sits on a chair. Its forearm is maintained on a table in front of his, with the elbow bent at 45° and the hand in the prono-supine position. MMG is recorded on the *flexor digitorum* muscle belly, either with the optical probe or with the microphone-based sensor.[10] Both signals are fed to a computer that displays in real time on a bar graph a pre-programmed consign level and the exerted force, so that the subject can regulate his effort.

For each subject, the maximum voluntary contraction (MVC) is first determined. That is the maximum force he can instantaneously exert under experimental conditions. It is used to normalize efforts for further comparison between subjects. This measurement is done three times, and the highest value is kept.

To compute the MMG amplitude versus force relationship, a sequence of five successive force levels of 15 s duration each is recorded. The force consign is equal to 15%, 25%, 30%, 35% and 45%, respectively, of MVC. For each stage, only the MMG corresponding to the stabilized force zone is retained.[21] That period, if it exists, is defined as the longer time interval, of at least 3 s duration, where the force never differs by more than 5% of the consign. The root mean square (RMS) value of the MMG is then

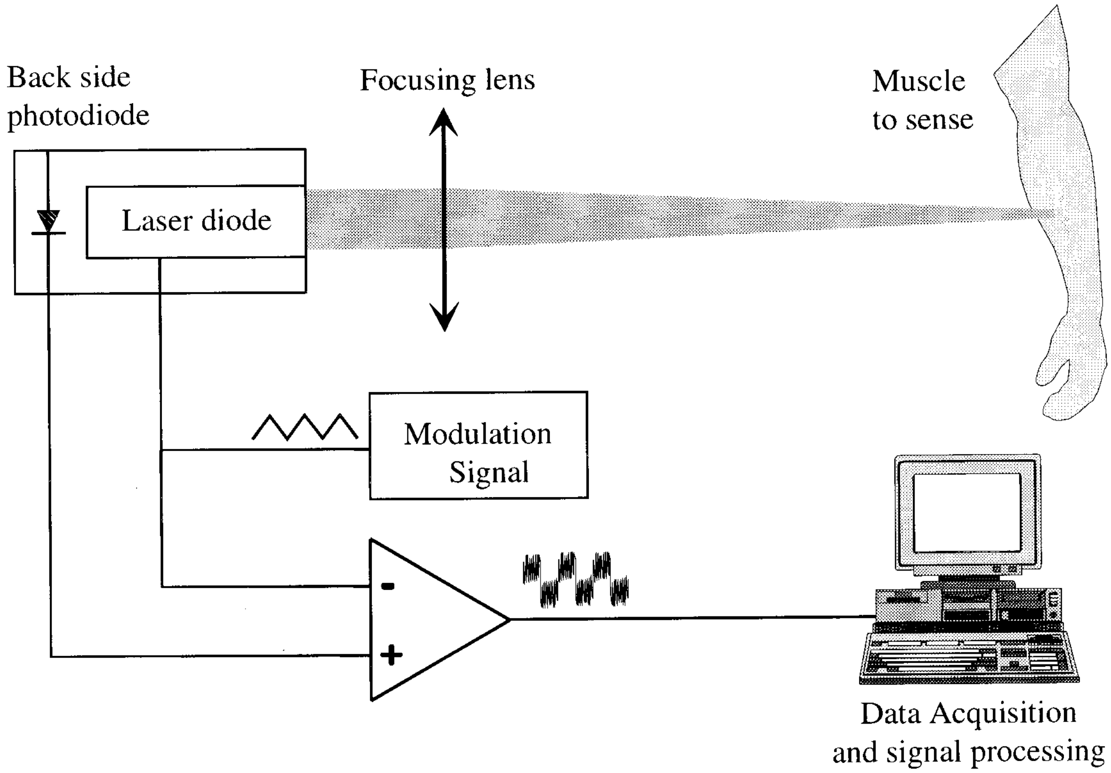

Fig. 1 Experimental setup.

computed from its spectral power density S in a frequency range $[f_{\mathrm{low}}, f_{\mathrm{high}}]$ using

$$V_{\mathrm{RMS}} = \sqrt{\sum_{f=f_{\mathrm{low}}}^{f_{\mathrm{high}}} S(f)}. \tag{4}$$

3.2 REFERENCE SENSOR

We use the sensor described by Courteville et al. in Ref. 10 as the system of reference for the optical probe characterization, because it measures the MMG in a classical way (by contact) and is calibrated in frequency and amplitude. Its principle is summarized hereafter. It features a membrane in contact with the skin, a closed air chamber, and a microphone. The skin vibration due to MMG modulates the acoustic pressure inside the chamber through the membrane, which is sensed by the microphone. In this setup, the acoustic pressure due to MMG is high enough so that the microphone works in its optimal dynamic range.

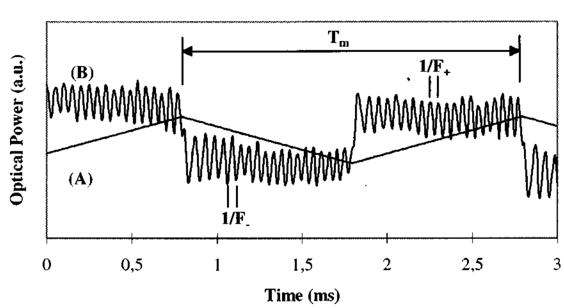

Fig. 2 Optical power P (curve A) and power variation due to feedback ΔP (curve B). The target velocity is computed through Eq. (3), using the ΔP frequencies F_+ and F_-.

4 MEASUREMENTS

We conclude from the experiment that the optical sensor works well when the skin surface is located within ±5 mm of the focusing lens focal plane. This condition ensures that there is enough power reflected and fed back into the laser cavity.

Figure 3 shows the frequency response (gain and phase) of the optical sensor and the reference acoustic probe. It was measured by means of a Burleigh piezoelectric pusher of 6 μm per 1000 V characteristic. A 2.8 μm p-p sinusoidal vibration was generated and measured directly at the ceramic pusher surface for 10 s by both sensors. The transfer functions were evaluated by computing the amplitude ratio and the phase difference between measured and expected vibrations. The procedure was repeated for several frequencies.

The spectral density of the distance equivalent noise is displayed in Figure 4 for the optical sensor. It is defined as the output noise when the sensor is focused on a motionless target. It was measured for 15 s. The distance equivalent noise, obtained by integrating the spectral power density under that curve, is 0.16 μm in the 1–100 Hz frequency range, and becomes 0.07 μm in the 5–100 Hz range.

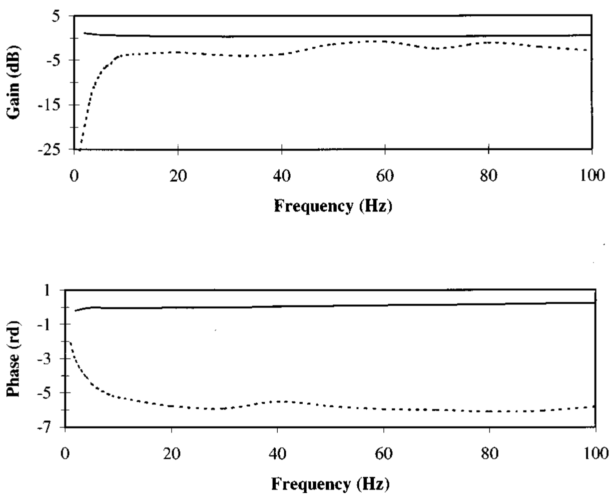

Fig. 3 Frequency responses of optical sensor (continuous line) and reference acoustic sensor (dashed line).

Figure 5 shows the MMG RMS value plotted against the force. The recordings were made according to the protocol described in Sec. 3. The MMG was measured, respectively, by both sensors, on two male subjects, ages 27 and 40 years, who volunteered to participate in this study. The MMG rms value corresponding to the stabilized force zones was computed by Eq. (4), in the 5–30 Hz frequency range.

We also investigated the influence of laser light penetration into skin by MMG measurement. Therefore, the MMG was first measured with the optical sensor focused directly on the skin. Then, the skin was coated with a reflecting powder to prevent light penetration, and new measurements were made. In both case, four stages of 15 s duration each with a force consign set to 30% MVC were recorded. For each recording, the power spectral density of the MMG corresponding to the stabilized force zone was computed. The MMG mean power spectral density, computed over the four stages and expressed with a confidence range at 90% using student law, is displayed in Figure 6 for both cases (with and without light penetration).

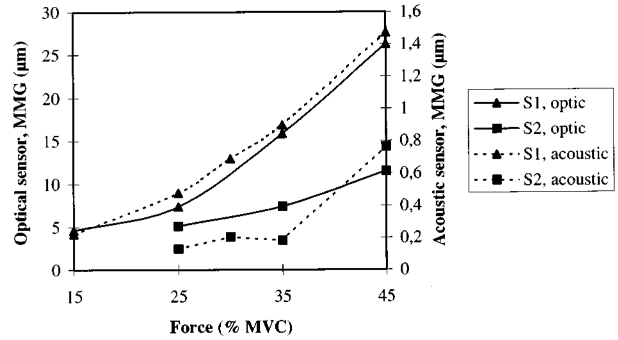

Fig. 5 MMG RMS value in the 5–30 Hz frequency range plotted against force, for two subjects (S1 and S2), and recorded, respectively, with the optical sensor (continuous curve, left *Y* scale) and the acoustic sensor (dashed curve, right *Y* scale).

5 DISCUSSION

This study describes an optical MMG sensor based on the feedback effect in a laser diode. As the optical setup requires few components and is self-aligning, so it is well adapted for clinical use. The probe shows innovative characteristics. Its transfer function, in phase as well as in amplitude, is flat over the whole frequency range (Figure 3). Furthermore, there is no low cutoff frequency and the upper spectral limit is set only by the modulation frequency through the sampling theorem. As the noise RMS value is about 100 nm (Figure 4), submicrometer vibrations can be measured. The sensor sensitivity is indeed high enough for MMG, even if it decreases under 5 Hz, as the noise increases.

Perhaps the main feature of this sensor is that the measurement is made without contact. MMG can therefore be sensed on nearly every superficial muscle, even the small ones, and on a precise place. Furthermore, the muscle perturbation is avoided. This last effect is clearly noticeable when comparing data from the acoustic probe and the optical one (Figures 3 and 5). The calibration curves (Figure 3)

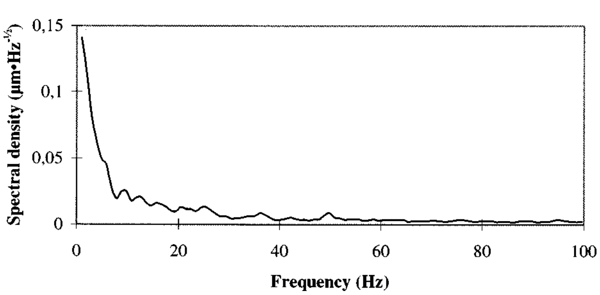

Fig. 4 Spectral density of the distance equivalent noise, computed with a 1 Hz resolution from 15 s of data. The laser beam is focused on a steady (motionless) target.

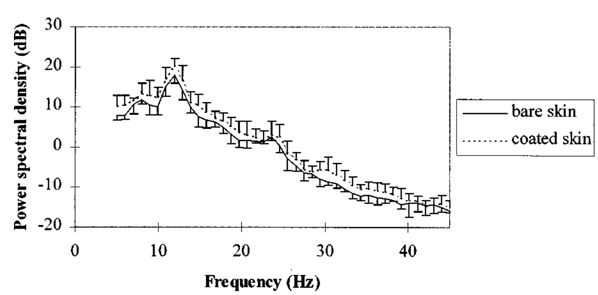

Fig. 6 MMG mean power spectral density sensed, respectively, on the bare skin or the reflecting powder coated skin, and computed with a confidence range at 90%.

show an amplitude ratio equal to 2 between the frequency responses of both sensors in the 10–40 Hz range. For MMG measurement, however (Figure 5), there is quite the same evolution against force with both sensors, for the two subjects, but the amplitude of the MMG sensed by the acoustic sensor is 20 times smaller than when sensed with the optical probe. The difference (a factor of 10) is to be attributed to the mechanical coupling between the muscle and the acoustic sensor membrane. That demonstrates the interest of a measurement without contact.

We have investigated the noise sources that can perturb the MMG measurement. The first one that can influence sensors such as microphones is the acoustic (ambient) noise. Fortunately, because of its principle, the optical probe is insensitive to it. Another noise source is physiological noise, such as heart beats, breathing, voice. As it is a vibration propagating into the body, it can be superimposed onto the MMG sensed by every kind of sensor. It is in fact the main noise source. The optical sensor is also sensitive to the movement due, for example, to the muscle contraction. It appears in the recordings as a high amplitude low frequency (≤ 1 Hz) signal. As it is not in the same frequency range as the MMG, it can be rejected by filtering.

We have also focused on a phenomenon that is specific to the optical method: the light penetration into skin. The light penetration depth depends on the wavelength, and is maximal between 600 and 1300 nm. De Mul et al.[22] have indeed measured subcutaneous blood flow using feedback in the laser diode. In our case, however, as Figure 6 shows, light penetration appears to be of little influence on the MMG spectrum.

6 CONCLUSIONS

We have described a MMG interferometric sensor based on the feedback effect in the laser diode. Its key features are

1. noncontact measurement, avoiding muscle perturbation;
2. measurement possible on nearly all superficial muscles;
3. extended bandwidth, with flat frequency characteristics;
4. high sensitivity;
5. simple, self-aligning setup, adapted to clinical use; and
6. immunity to ambient noise.

For this sensor, the main noise source is the physiological noise. It is also sensitive to the global movements, which fortunately can be filtered from the MMG signal. We have also demonstrated the influence of the contact, inherent to every classical kind of MMG sensor.

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