Towards an acoustic map of abdominal activity

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Abstract— **The general framework of this communication is phonoenterography. The ultimate goal is the development of a clinical diagnostic tool based on multi-channel abdominal sound monitoring. This paper concerns essentially the source localization problem in phonoenterography. After studying different sound localization methods, all based on abdominal surface auscultation, we propose and justify our method and we present the first results we have obtained on real multi-channel recordings.**

Keywords— **Bowel sounds, phonoenterography, source localization, sound propagation.**

I. Introduction

Bowel sounds (BS) have been attracting attention for a long time, since they possibly carry functional information about the digestive tract. Different normal or pathological physiological phenomena, varying in time according to the different phases of the digestive cycle and occurring in different abdominal locations can thus be studied by means of BS recording, preprocessing and interpretation. We have presented our preprocessing method (wavelet denoising and segmentation) in [1]–[3].

Following steps concern the extraction and the interpretation of the information embedded in the signals. This information is generally distributed upon long periods of time: digestive phenomena are described using activity indexes that gather statistic information as the time distribution of the BS (number of sounds/minute), the sound-to-silence ratio, the main-frequency distribution, the energy distribution, etc. [4]–[8], but seldom the spatial distribution of gastro-intestinal activity, even if it might be carrying important information on the abdominal physiology. The goal of this communication is to study the bowel sounds localization methods described in different works, in order to help choosing the most reliable one and to assess its usefulness in phonoenterography.

II. Localization: a state of the art

Bowel sounds source localization is seldom performed. The main cause seems to be the ambiguous information offered by abdominal auscultation: bowel sounds are often propagated and heard at any location, making localization attempts difficult or useless. Several authors detect time variation of gastric activity, but no or insignificant location differences [4], [9], [10]. Still, several authors report a correlation between certain locations or even certain diseases and types of sounds and/or activity [11]–[13]. To our knowledge, the most advanced works on BS localization are:

- Craine *et al.* [13], which uses 3 electronic stethoscopes placed on the abdominal wall and an isotrope model of the propagation environment, and reports differences in the localization of the small bowel activity for several functional diseases (Crohn's disease, Irritable Bowel Syndrome – IBS);
- Garner and Ehrenreich [14] (5 microphones), which suppose a sound propagation based on the anatomy of the abdomen and on medical expertise (BS propagate along the intestines). Still, their method lacks of systematic validation, as this anatomic propagation model cannot take into account the variations of the abdominal status (e.g. bowel content variations during digestive cycle or between patients).

In fact, any localization attempt implicitly assumes a model of sound propagation. The most common and the simplest one is to consider that the sound source is closer to the microphone that recorded the strongest signal [11], [14].

In [14], authors also suggest that bowel sounds can propagate along the intestines. Recently, the localization method presented in [13] uses an isotrope model for the abdomen and propose a source triangulation using an non-absorbent propagation model.

Neither one of these methods is thoroughly validated: the goal of this work is to study the different models and to propose the most reliable one.

III. Proposed localization methods

A. Propagation models

For most applications, localization is based on the evaluation of the time-of-arrival (TOA, time difference between the arrival of a sound at different locations). For BS localization, the authors of [14] propose a TOA method, under the assumption that guts are empty cavities and the sound speed is the same as in air (≈ 340 m/s). This hypothesis is hard to sustain in all physiological situations, knowing also that the measured sound speed in human tissues is rather close to the sound speed in water (≈ 1500 m/s) [15]. Moreover, this propagation speed allows us to affirm that TOA methods are

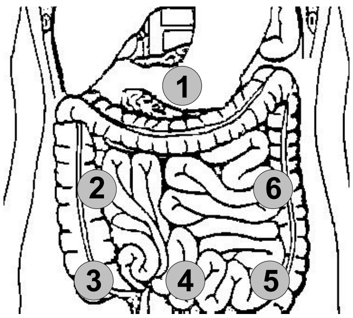

Fig. 1. Stethoscope placement.

hard to implement, as distances between different possible source locations and microphones are too small.

A more formalized method is the sound intensity based triangulation, which assumes in fact an isotrope environment. Two propagation models can be used: absorbent and non-absorbent.

1) Non-absorbent model: The authors of [13] consider that the intensity of a BS decreases with the square of the distance between a *point source* and the microphone (inverse square law [16]), which writes in a 3-dimensional Euclidean space:

$$I_i = k\frac{P_s}{4\pi d_i^2}, \tag{1}$$

with $d_i = \sqrt{(x_i - x_s)^2 + (y_i - y_s)^2 + (z_i - z_s)^2}$ the distance between the source and the microphone i, I_i the intensity of the sound as recorded by the microphone i, P_s the source power, and k a proportionality constant depending on the environment. If we consider the classical propagation model in a non-absorbent isotrope environment [16], $k = 1$, but the authors of [13] give no indication on its value. We can observe that, for $k < 1$, the environment becomes absorbent, but the absorbed energy does not depend on the distance between the source and the microphone.

2) Absorbent model: Another possible model is to consider an absorbent environment model [15] applied to the abdomen. The intensity of a propagated recorded sound writes then:

$$I_i = P_s\frac{e^{-\mu d_i}}{4\pi d_i^2}, \tag{2}$$

with μ the absorption parameter depending on the environment but also on the frequency of the propagated sound.

The authors of [13] use three microphones placed at the vertices of an equilateral triangle, so a unique solution of the triangulation problem always exists. Still, as the implicit hypothesis is that propagation takes place in an isotrope environment, the solution will be false if the hypothesis is wrong. The truthfulness of the localization cannot be assessed because of the unique solution of the triangulation, which offer no redundancy. Our first attempt was to verify the isotrope hypothesis using more microphones (the 6 stethoscopes we used were placed according to Fig. 1).

B. Localization cost-function

Consider $P_{s,i}$ *the estimate* of the power of the source placed at (x_s, y_s, z_s) computed from the intensity I_i of the sound recorded by microphone i, placed at (x_i, y_i, z_i) (with a constant z_i, as the stethoscopes were placed upon the abdominal surface, considered plane). In an isotrope environment, the source powers computed with equations (1) or (2) are equal: $P_{s,i} = P_{s,j}, \forall i, j$. We can then localize the sound source by minimizing the following cost-function:

$$\mathcal{F}(x, y, z) = \sum_{\substack{i,j=1 \\ i \neq j}}^{6} (P_{s,i} - P_{s,j})^2, \tag{3}$$

which verifies that $P_{s,i} = P_{s,j}$ for all permutations (i, j).

The coordinates (x_s, y_s, z_s) of the cost-function minimum give us the location of the BS source. One can check that in an ideal isotrope environment and considering no measure errors, $min\,(\mathcal{F}) = 0$. Still, as generally $min\,(\mathcal{F}) \neq 0$, the cost-function can be seen as a "measure" of the difference between the estimates of the source power. Therefore, its minimum value evaluates either the accuracy of the chosen model of the environment, or the instrumentation noise. As the measure errors can be easily minimized by carefully calibrating the acquisition chain, one can consider that the value of $min\,(\mathcal{F})$ is reflecting mainly the modelling error.

We have tried to localize real bowel sounds using the cost-function (3), using both models described above, non-absorbent and absorbent isotrope environment.

1) Non-absorbent environment: The cost function writes:

$$\mathcal{F}(x, y, z) = \left(\frac{4\pi}{k}\right)^2 \sum_{\substack{i,j=1 \\ i \neq j}}^{6} (I_i d_i^2 - I_j d_j^2)^2. \tag{4}$$

The first important observation we can make is that the solution of the optimization problem (the source location) does not depend on the absorption constant: introducing k in equation (1) does not change the localization, so one can choose the theoretical value $k = 1$[1].

2) Absorbent environment: The absorption constant μ introduced in equation (2) depends on the propagation environment and on the frequency of the propagated wave:

$$\mu = \alpha \cdot \nu^{\beta}, \tag{5}$$

with ν the frequency of the sound, α and β environment-specific constants. For soft human tissues, the usual value of β is 1 and the mean value of α is $8.3 \cdot 10^{-6}$ (s/m) (see [15] for more detailed information about particular tissues).

It is obvious that for $\alpha = 0$, the absorbent and non-absorbent models are identic, so we will consider only the absorbent one in the following and we will modify α in order to check both models performances.

[1]If the absorption constant $k \neq 1$ in equation (1), the value of $\mathcal{F}$ depends on it and it must be normalized (multiplied by k^2) before directly considering it as a measure of accuracy.

C. Model testing

In order to apply the previously described localization method, we need to calculate the intensities I_i and the principal frequency f of each bowel sound:

- the intensity of a sound was defined as the amplitude of its most energetic part;
- the principal frequency was computed as the peak of the Fourier transform of this most energetic part[2].

As a result of the previous preprocessing steps [1]–[3], each sound is characterized by its discrete wavelet decomposition coefficients. This representation allows us to extract both the intensity and the pitch of a sound.

1) Intensity: We have chosen to characterize the time distribution of the energy of a sound by computing its power upon time-sliding windows. The size of a window is given by the length of the largest scale wavelet, and the sliding step by its position[3]. The square-root of a window-power is then a good estimate of the intensity of a BS. As the chosen wavelet decomposition is orthogonal, this value can be computed directly from the wavelet coefficients. For a given time position p (which corresponds to the p^{th} coefficient on the largest scale), the segmentation performed directly on the wavelet coefficients vector allows us to write the energy of the p^{th} window of the sound recorded by the i^{th} stethoscope:

$$E_{p,i} = C_{p,0}^2 + \sum_{j=1}^{M} \sum_{\rho=p-\tau}^{p} C_{\rho,j}^2, \quad \text{with } \tau = 2^{j-1} - 1.$$

The total energy of the recorded sound is $E_i = \sum_p E_{p,i}$, and its intensity is defined like:

$$I_{i,p_{max}} = \max_p \sqrt{E_{p,i}}, \tag{6}$$

with p_{max} the time-position of the most energetic window.

2) Main frequency: As both the discrete wavelet decomposition and the Fourier series decomposition are linear, the Fourier transform of a signal (which is represented as a sum of wavelets ψ and scaling-functions ϕ) can be calculated as the sum of the Fourier transforms of its wavelets. Using a simplified notation, this observation writes:

$$s(n) = \sum_p C_{p,0}\phi_{p,0} + \sum_{p,j} C_{p,j}\psi_{p,j} \Rightarrow$$
$$\widehat{S}(f) = \sum_p C_{p,0}\widehat{\Phi}_{p,0} + \sum_{p,j} C_{p,j}\widehat{\Psi}_{p,j},$$

were $p > 0$ represents the wavelet position in time and $j = 1..M$ (M being the depth of the decomposition) the scale, $C_{p,j}$ its amplitude and $\widehat{\cdot}$ the Fourier transform. The main frequency of the signal can then be computed as the peak of the sum of the Fourier transforms of the wavelets that compose it:

$$\nu = \arg\max_f \left|\widehat{S}(f)\right|.$$

The same argument can be applied to a part of a signal, which in our case is the most energetic window p_{max}. The principal frequency of a bowel sound recorded by the i^{th} stethoscope writes then:

$$\nu_{i,p_{max}} = \arg\max_f \left|\widehat{S}_{i,p_{max}}(f)\right|. \tag{7}$$

3) Cost-function optimization: The cost-function (3) is a hyper-parabola which has a unique global-minima. The coordinates of this minima (x_s, y_s, z_s) represent the estimated position of the sound source and allow us to calculate the distances between the source and every stethoscope. Introducing this distances in equation (2), we obtain, for each stethoscope i, an estimate $P_{s,i}$ of the power of the source. The sum of quadratic differences between all pairs of estimates $P_{s,i}$ is in fact the cost function (3), which minima can then be used as a measure of the validity of estimates $P_{s,i}$ and thus of the chosen model for the localization.

In fact, the minimum value of the cost-function increases with the spread of $P_{s,i}$, but it depends also on the actual values of $P_{s,i}$ (related to the power of the sound) and on the number of channels on which the actual sound was recorded. Therefore, we have considered only the spread of $P_{s,i}$ estimates in order to evaluate the accuracy of the model. A good measure of the this spread can then be the mean of the normalized quadratic relative error:

$$\epsilon^2 = \frac{1}{N}\sum_{i=1}^{N} \varepsilon_i^2 = \frac{1}{N}\sum_{i=1}^{N} \left(\frac{P_{s,i} - \frac{1}{N}\sum_{i=1}^{N} P_{s,i}}{\frac{1}{N}\sum_{i=1}^{N} P_{s,i}} \right)^2, \tag{8}$$

with N the number of stethoscopes. An $\epsilon \approx 0$ means that $P_{s,i}$ values are close (and close to their mean value), so the cost-function has a small minima and the model is validated.

IV. Results

Bowel sounds were recorded using $N = 6$ stethoscopes placed on the abdominal surface like in Fig. 1. After anti-aliasing low-pass filtering (2250 Hz.), the signals were digitized at a sampling frequency of 5000 Hz. The duration of the recordings was approximately 7 minutes (2^{21} samples). Wavelet denoising and segmentation were performed on the 6 channels. The segmented events were considered as generated by a unique propagated bowel sound if they were overlaping in time on at least 2 channels.

We have tried to localize bowel sounds using both models described above. Still, for the human soft tissues and for the low frequencies [100–1000] Hz. of physiological sounds, the absorption constant $\mu \approx [8.3 \cdot 10^{-4} - 8.3 \cdot 10^{-3}\ \text{m}^{-1}]$ (5). This means that, for small distances (as in BS propagation), the absorbent model (2) will be almost equivalent to the

[2] This choice was made in order to ensure the coherence of the propagation model, but also because we considered that it is the most representative for the auditive impression (pitch).

[3] These parameters are function of the chosen wavelet transform, which in our case is an orthogonal decomposition on Daubechies-9 wavelet-basis.

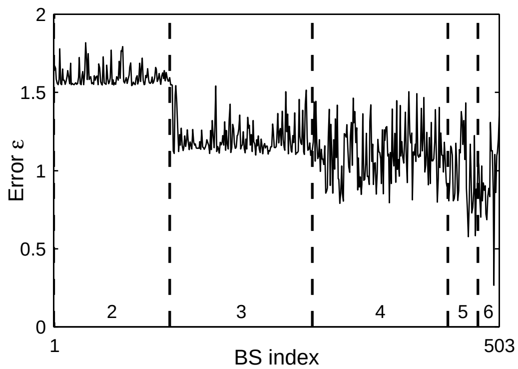

Fig. 2. Error ϵ for the 503 bowel sounds. Its value depends on the number of stethoscopes that recorded the sounds (2 to 6, printed in the low part of the graphic). As expected, the isotrope environment localization model is more accurate for the sounds that reach a greater number of stethoscopes.

non-absorbent one (1). The localization was performed by minimizing the cost-function (3) for all sounds recorded on at least 2 channels. During the 7 minutes of recording, we have detected 738 bowel sounds, with 503 BS appearing on more than 1 channel. For the 503 sounds we have computed the location of the source and the error ϵ.

As expected, the two models gave approximately the same results both in terms of localization and error. Nevertheless, these results were disappointing: the mean value of $\epsilon \approx 1.25$ ($0.27 \leq \epsilon \leq 1.82$). Localization is thus highly inaccurate: the difference between source intensity estimates is greater than their mean value (see Fig. 2).

On the other hand, anatomic knowledge, as well as ultrasound applications, suggests that the absorbent (but anisotrope) environment hypothesis is correct. In this case, one can consider a less restrictive model of propagation, which writes, using the same notations as in (1) and (2):

$$I_i = P_s f(d_i), \tag{9}$$

with f a monotonic decreasing function.

As BS intensity diminishes with distance, the source is closer to the stethoscope that recorded the strongest signal[4]. Under this hypothesis, one can locate a BS by assigning it to a single stethoscope and delete it on the other channels. For the 7 minutes recording, the results presented in table I show important changes in the evaluation of the abdominal activity when localization is considered.

V. Conclusion and Perspectives

The previously developed approach allows us to conclude that triangulation in isotrope environment cannot be used for BS source localization, as the modelling error is too important. The simplest and the most widely used localization method, which assigns each sound to one stethoscope, is

TABLE I

No. of BS and total energy before and after localization

Stethoscope	Before		After	
	No. BS	Energy	No. BS	Energy
1	55	1 837	11	990
2	555	13 996	61	1 198
3	795	56 294	538	51 557
4	439	22 366	90	8 714
5	240	8 504	23	1 117
6	73	3 857	15	2 623

then the most precise, at least until further development of acoustic models of the abdomen. Using this approach, we are currently conducting studies on healthy volunteers, upon several hours of recording. Preliminary results (not detailed in this work), show location variations of the abdominal activity during the digestive cycle.

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