# Design limitations of using magnetic inductive coupling to verify nasogastric tube placement


Nuno P. Silva*, Adnan Elahi*, Martin O'Halloran*, Derek T. O'Keeffe† Declan O'Loughlin‡§

*Translational Medical Device Lab, University of Galway, Galway, Ireland

†Division of Endocrinology, Department of Medicine, University of Galway, Galway, Ireland

‡Electronic and Electrical Engineering, Trinity College Dublin, Dublin, Ireland

§Trinity Centre for Biomedical Engineering, Trinity College Dublin, Dublin, Ireland

n.silva1@universityofgalway.ie



***Abstract*—Nasogastric tube insertion is a standard procedure but with potentially fatal consequences if misplaced. Many bedside methods have been proposed for identifying misplacement including magnetic inductive coupling. This study aims to identify the coil design limitations for nasogastric tube placement confirmation using magnetic inductive coupling. The fundamental limits on the coil sizes and detection distances are identified based on the anatomical understanding. Candidate designs are simulated and fabricated, and key parameters such as coil radius and penetration depth are examined. Specifically, the mutual inductance and coupling factor between the coils in the candidate designs are both calculated and measured with respected to distance. This work highlights that a small coil of up to 3 mm in diameter could be integrated into a nasogastric tube without impairing clinical function. These preliminary results suggest that magnetic inductive coupling is a promising approach for nasogastric tube localisation, however, optimisation of the detector coil size could improve localisation performance.**



***Index Terms*—electromagnetic coupling, electromagnetic induction, inductive power transmission, medical devices and systems, mutual coupling.**


## I. Introduction

Many patients who have difficulty swallowing or consuming nutrition via the mouth will have a nasogastric tube inserted to provide continuous feeding in the short-term [1]. The nasogastric tube can be used to provide nutrition, hydration and other oral agents among others, but also for gastric decompression or irrigation [2], [3]. In emergency situations, or for short-term use, nasogastric tubes are a quick, easy and relatively non-invasive medical procedure to prevent malnutrition or dehydration. However, great care needs to be taken to ensure the nasogastric tube does not lodge into the lungs, trachea or oesophagus during insertion. In these cases, severe injuries, such as pneumonia, pulmonary haemorrhage, subsequent liquid feeding or even death can occur due to the misplacement of the nasogastric tube. Nevertheless, studies in the UK suggest that approximately 3–4 % of nasogastric tubes are misplaced in the respiratory system and 19 % on the oesophagus each year in the UK, with potentially fatal consequences [2].

In the attempt to avoid nasogastric tube misplacement and any other complications, guidelines have been developed to allow the position of the nasogastric tube to be confirmed. It is recommended using pH testing and/or a chest X-ray to confirm the nasogastric tube placement [1], [4], [5]. Although a positive development, studies have indicated that the false positive rate of these tests are quite high, as many severe injuries and deaths due to a misplacement of the nasogastric tube are still reported [6], [7], [8]. Despite the inclusion of chest X-ray in the guidelines, misinterpretation of the chest X-ray accounts for many of the serious incidents reported in the UK [1].

Magnetic tracking and electromagnetic induction have been proposed as alternatives which are easy-to-use, low-cost and safe. These methods involve integrating a permanent inductive electromagnetic coil into the tip of the nasogastric tube to track the location of the tube. These approaches have a number of advantages, specifically:

1) the magnitude of the magnetic susceptibility of biological tissues is very small and the effects are negligible at the field strengths needed for detection;
2) the organs of interest (namely the stomach, lungs and oesophagus) do not typically overlap anatomically from the perspective of the handheld transmitter, improving the reliability of the method (similar to ultrasonography);
3) integration of the required magnet or coil into the nasogastric tube is feasible;
4) and finally, the device would be easy to use and interpret, and not require substantial retraining of the user.

This work examines the design limitations of an inductive, electromagnetic coil for nasogastric tube localisation at the patient-bedside. Specifically, the fundamental limits due to the size and nature of the nasogastric tube are identified and the limits on the integrated coil size determined. Based on these limitations, the implications for the size of the detector coil are examined in light of typical anatomy with the tube inserted.

Similar magnetic inductive coupling technologies are widely used in biomedical applications such as wireless charging [9], [10]. Here, a receiver coil is integrated into the tip of the nasogastric tube designed to couple with a transmitter coil integrated into a handheld transmitter. A known magnetic field is generated by the transmitter coil, and the power lost in the transmitter coil is monitored. When the transmitter coil in the handheld transmitter and the receiver coil in the nasogastric tube are overlapping, the magnetic field induced by the transmitter coil induces a small current in the receiver coil. This induced current causes power to be transferred from the transmitter coil in the handheld transmitter to the receiver coil, located in the nasogastric tube tip. This causes power loss in the transmitter coil which can be quantified and measured.

Therefore, this work looks at some specific technical factors considering the design limitations of the coils:

- firstly, the receiver coil inserted into the nasogastric tube tip is designed such that the nasogastric tube functionality is not impaired and the physical dimensions of the tube are not substantially altered;
- secondly, the mutual inductance and coupling factor between the receiver coil in the tip of the nasogastric tube and the transmitter coil radius in the handheld transmitter are simulated and measured experimentally for a range of distances between both coils;
- finally, the effect of the coil size on the mutual inductance and coupling factor between the coils is investigated.

Together, these results help establish the technical requirements for successful implementation of a nasogastric tube detector in terms of required coil radii to ensure detection at realistic distances.

The remainder of this paper is structured as follows: Section II discusses the background to the nasogastric tube and the methods used in the paper. Section III presents the results and finally, Section IV concludes the paper.

## II. Methods

A nasogastric tube is flexible plastic tube with a minimum diameter of 3 mm, as seen in Fig. 1. The tube is inserted through the nose and fed until it is in the stomach.

Although, the tube is flexible, the tip of the tube is more rigid to assist in insertion, allowing for the receiver coil and other circuitry to be integrated into the tube without impairing the clinical functionality. The tube is open-ended right before the tip to allow for the nutrition to pass through, allowing for the coil to fit in the tip without interfering with the functioning of the nasogastric tube. However, the size of the nasogastric tube means the design parameters of the receiver coil are fundamentally limited. Most importantly, the receiver coil must be, at most, 3 mm in diameter to ensure the size of the nasogastric tube is not changed. The coil may be integrated into the tip of the tube by the manufacturer, or be fitted to the tube separately.

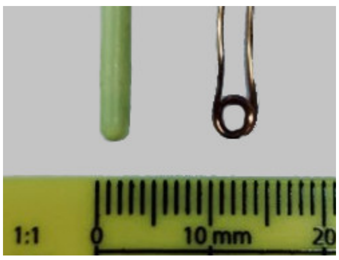


Fig. 1. Nasogastric tube tip and coil prototype.

On the other hand, the transmitter coil radius can be freely chosen in order to maximise the detection of the receiver coil, and consequently the nasogastric tube placement. Fundamentally, high coupling is desired across the range of distances required, as higher power transfer will be easier to detect in the handheld transmitter.

### A. Electromagnetic detection

Considering two ideal coils in free space, the coupled power between the two coils can be maximised if both coils are designed to resonate at the same resonant frequency. When the resonant frequencies of both coils are the same, the maximum power is transferred. For both coils, a capacitor is included in series with the coil to minimise the magnitude of the reactance and control the resonant frequency, to maximise the power transferred between both coils.

The circuit is shown in Fig. 2. The transmitter coil in the handheld transmitter is shown on the left, excited at the resonant frequency of the receiver coil that is shown on the right. The circular coils were fabricated using an electrically-conducting copper wire into $N$ turns, with a cross-sectional radius, $r$ [m], and a total coil length, $l$ [m]. The inductance of the coil, $L$ [H], is then calculated analytically using:

$$L = \frac{\mu_0 N^2 \pi r^2}{l}, \tag{1}$$

where $\mu_0$ is the magnetic permeability of free space equal to $4\pi$ x $10^{-7}$ H m$^{-1}$.

Then, both $LC$ circuits (in the nasogastric tube tip and in the handheld receiver) were optimised to the same resonant frequency by the inclusion of a capacitance, $C$:

$$f_r = \frac{1}{2\pi\sqrt{LC}}. \tag{2}$$

Hence, power transfer between the two coils was maximised, facilitating detection.

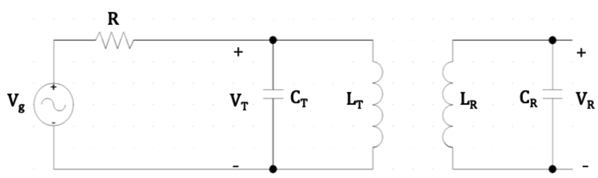


Fig. 2. Circuit diagram of the proposed solution to nasogastric tube detection. The circuit on the left is integrated into a handheld receiver which is controlled by the healthcare professional, whereas the circuit on the right is integrated into the tip of the nasogastric tube. The tube detection can be determined by a small decrease of $V_T$, when the transmitter coil ,$L_T$, is excited and closed to the receiver coil, $L_R$ In this case, the transmitter coil induces a small current in the receiver coil that can be measured by the decrease in voltage at the terminals of the transmitter coil.

### B. Mutual inductance experimental set-up

Mutual inductance is a quantitative measure of the coupling between two adjacent coils: *i.e.* mutual inductance is a measure of the ability of the transmitter coil to induce a current in the receiver coil. As maximising power transferred will maximise the distance at which the coil can be detected, understanding the limits on mutual inductance give the physical size of the nasogastric tube will help demonstrate the possible usage of this approach.

A number of techniques exist for estimating the mutual inductance between two coils. A common approach suitable for this application is the voltage-ratio method, that exploits the measurement of the voltage in the transmitter coil, $V_T$ [V] and the receiver coil $V_R$ [V] during excitation of the transmitter coil [11]. The mutual inductance between both coils, $M$ [H], can be determined by the relationship between both voltages and the self-inductance of the transmitter coil, $L_T$ [H]:

$$M = \frac{L_T V_R}{V_T} \tag{3}$$

The coupling factor, $k$, is then determined by the ratio between the mutual inductance and the self-inductance of the coils:

$$k = \frac{M}{\sqrt{L_T L_R}} \tag{4}$$

where the coupling factor is a measure of the power transferred between the coils during excitation.

To estimate the mutual inductance between two coils, electromagnetic simulations were conducted using COMSOL, Stockholm, Sweden. All coils were simulated using copper wire of 0.5 mm diameter, and the transmitter coil was excited with a sinusoidal stimulus of amplitude 5 V peak-to-peak and a frequency of 2 MHz. Receiver coils of diameter 3 and 5 mm was simulated with transmitter coil diameters of 25, 50 and 100 mm and 10 turns.

To validate the simulations and investigate the design limitations of this approach in realistic, experimentally, a receiver coil was fabricated at the same dimensions as the simulation. Similarly, a transmitter coil was fabricated with diameter of 50 mm and 10 turns. The self-inductance of each coil was calculated analytically using equation 1 and also measured with an LCR meter (LCR200, EXTECH Instruments, Waltham, MA, USA) which had a resolution of 5%.

Capacitors were chosen to give each circuit a resonant frequency of 2 MHz using (2), as in other studies [12]. A number of factors influence the frequency choice in broad terms:

- firstly, as the inductance of the receiver coil is limited by the physical constraints of the nasogastric tube: lower frequencies require smaller integrated capacitors;
- secondly, at higher frequencies, eddy currents in the transmitter coil limit the potential power transmitted and increase the power losses in the transmitting coil itself not due to the presence of the receiving coil;

However, in the low MHz range, initial investigations suggest that these factors do not significantly impact the detection. In the the chosen frequency range, absorption of the electromagnetic energy from the body is low [13]. Additionally, the range and the purpose of the technology to be used in few centimetres between coils, does not require a specific frequency if it is set in low MHz [14]. A 1 kΩ resistor was using with the transmitter coil which was then excited using a signal generator (Agilent 33120, Santa Clara, CA, USA) to generate a sine wave with 2 MHz of frequency and 5 V of peak-to-peak amplitude measured in $V_T$, as shown in Fig. 2.

The voltages were measured by a multi-function instrument that can measure and record signals (Analog Discovery, DIGILENT, Austin, TX, USA). The uncertainty related to the measurements were determined by the difference between the minimum and maximum values obtained for 5 measurements: $\Delta V$ of 5 %. The uncertainty related to the inductance measurements is determined as the minimum resolution of the device. Then, mutual inductance and coupling factor uncertainties were calculated accordingly to [15].

## III. Results and Discussion

The results from experimental and simulation are presented in this section aside with discussion.

### A. *Proof-of-concept design*

The proof-of-concept experimental set up was designed in order to measure the mutual inductance and the coupling factor of the transmitter and receiver coils as a function of the distance between them. The transmitter and receiver coils were fabricated with the specifications described in section II-B: a receiver coil diameter of 5 mm and a transmitter coil of 50 mm and 10 turns. The self-inductance of both coils was measured using a LCR meter ($L_T$ = 24.2 ± 0.1 μH and $L_R$ = 2.1 ± 0.1 μH) and calculated analytically using (1): $L_T$ = 25.9 μH and $L_R$ = 2.6 μH. The measured and calculated self-inductances have an error percentage of approximately 7 and 19%. Errors between the measured and analytical values occur because of errors in the fabrication process, specifically, separation between the turns, resistance of the wire and other imperfections.

The measured values for mutual inductance were calculated measuring the voltage at the terminals of each coil ($V_T$ and $V_R$, as seen in Fig. 2); and comparing the voltage ratios from (3). The experimental measurements of mutual inductance are presented in Fig. 3. Voltage measurements were conducted with both coils concentric but with separation between the planes of the coils. The receiver coil was then moved perpendicular to the coil. All experimental measurements were conducted in the same conditions twice, in order to observe the repeatability of the experimental procedure.

Additionally, the same scenario was simulated in COMSOL to validate the results. As observed in Fig. 3 the experimental mutual inductance is lower than the simulated mutual inductance, particularly when the coils are closer together. These differences would appear to be due to experimental error such as movement during the experimental measurement, imprecise alignment of the coils, and errors in the fabrication process. Based on the mutual inductance, the coupling factor was calculated by (4). Fig. 4 shows the determined values obtained in function of the distance between the two coils.

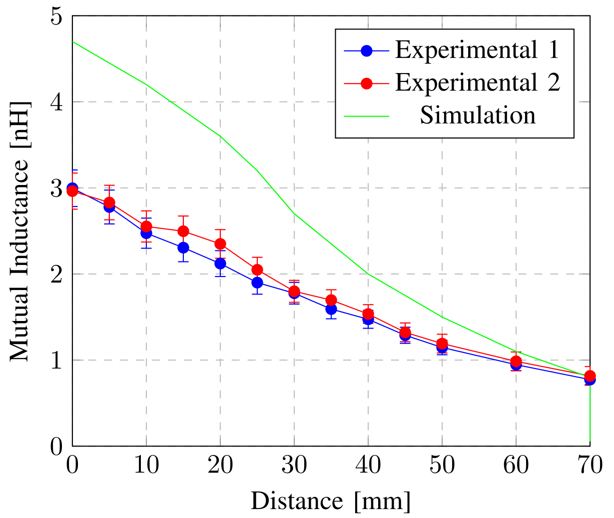


Fig. 3. Experimental and simulated mutual inductance in function of distance of a receiver coil diameter of 5 mm and a transmitter coil diameter of 50 mm. The vertical error bars represent the uncertainty associated with the measurement.

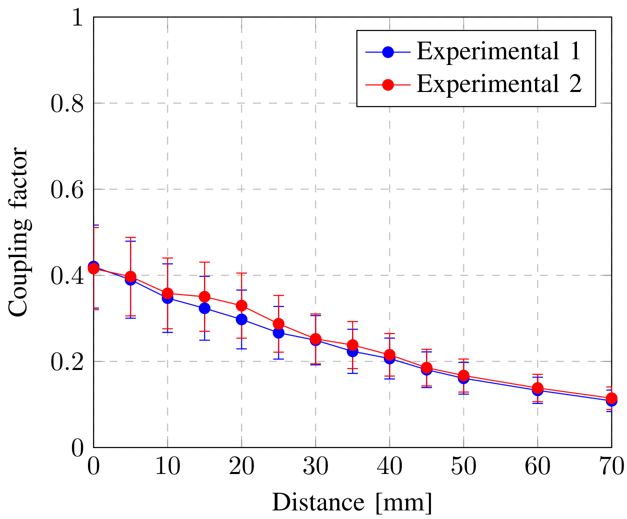


Fig. 4. Determined coupling factor in function of the distance between the transmitter and receiver coil with diameters 50 and 5 mm, respectively. The vertical error bars represent the uncertainty associated with the obtained values.

Considering both Fig. 3 and Fig. 4, it is observed that at close distances between both coils, the mutual inductance, and consequently the coupling factor is higher than when the coils are farther apart, which is expected. Additionally, the experimental and simulated results show the same trends, which validates the simulations although discrepancies do exist particularly at very close distances.

## *B. Transmitter coil design*

Initial simulations investigating the optimal size of the transmitter coil are conducted. In terms of practicality for use, if the transmitter coil is too small, the area of detection will be too small and it could be difficult to find the tip while scanning the abdomen. If the transmitter coil is too large, the area of detection will be too big and the likelihood of false detection is larger.

The simulations in this section look at the mutual inductance as a function of transmitter coil dimensions and inter-coil distance using the same simulation set-up as described in section II-B. The simulations were repeated for inter-coil distances of 25, 50, 75 and 100 mm, for different transmitter coil size. In all simulated distances and coil radius, the transmitter coil was induced with 5 V and frequency of 2 MHz.

The results obtained from simulation are showed in Fig. 5. As observed in Fig. 5, when there is an overlapping of the passive and transmitter coils (distance equal to 0 mm), the mutual inductance obtained is significantly higher than the mutual inductance obtained for the other distances between coils, for a transmitter coil radius between 5 and 50 mm. Afterwards, the mutual inductance obtained for the a transmitter coil radius higher than 50 mm has approximate values compared with the other distances between the coils.

The mutual inductance obtained in the simulation for distances of 25, 50, 75 and 100 mm have a similar behaviour. The mutual inductance first increases as the transmitter coil radius increases, until an transmitter coil radius of 40 mm and 80 mm in the case of coils separated by 25 and 50 mm, respectively. As the separation between the coils increases, the mutual inductance decreases.

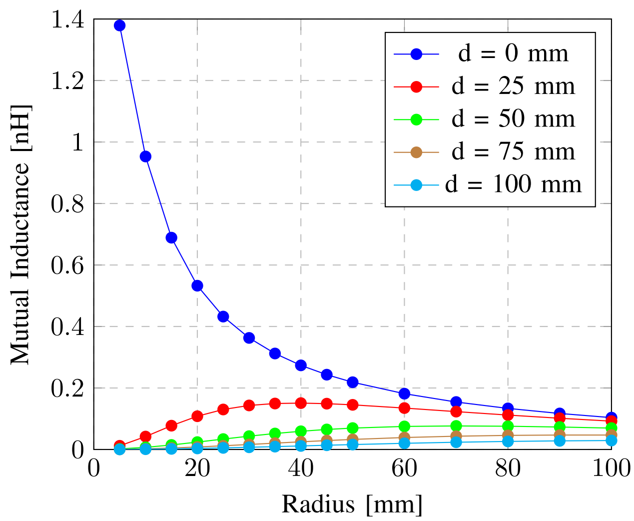


Fig. 5. Simulation of the mutual inductance in function of the transmitter coil radius, considering a receiver coil diameter of 3 mm separated from the transmitter coil by distances of 0, 25, 50, 75 and 100 mm.

In order to understand the depth in mutual inductance, Fig. 6 shows the mutual inductance as a function of the distance between the coils, considering a transmitter coil radius of 25, 50 and 100 mm. Fig. 6 demonstrates that care needs to be taken in the design of the transmitter coil. While a smaller coil transmitter coil leads to larger mutual inductances when the inter-coil distance is low (below 20 mm) when compared to the larger transmitter coils (50 mm and 100 mm), this trend

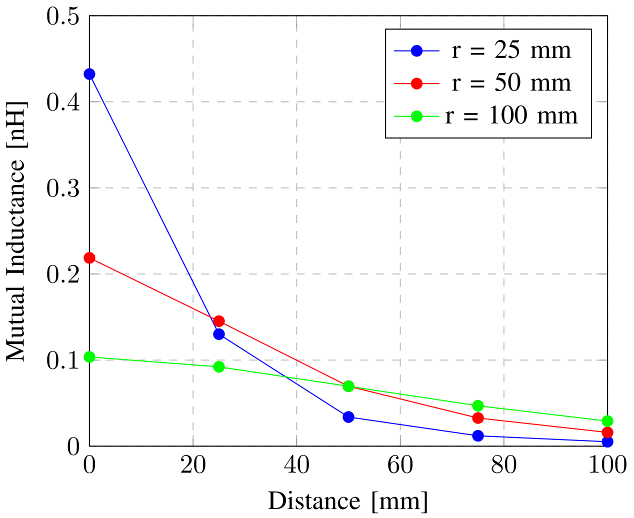


Fig. 6. Experimental evaluation of the mutual inductance in function of the distance of the transmitter and receiver coils, considering a receiver coil diameter of 3 mm and three different transmitter coil radius - 25, 50 and 100 mm.

is reversed at higher intercoil distances. Fig. 6 shows that depending on the expected intercoil distances, the outer coil radius can be optimised to maximise detection.

Considering the specific applicability of the electromagnetic induction in order to detect the tip of the nasogastric tube, it is important to take in consideration the limitation or necessities related with the detection: the receiver coil, that will be inserted into the nasogastric tube tip has limit size dimensions. However, the transmitter coil — detector coil — can be dimensioned in order to maximise the mutual inductance, *i.e.* maximise the magnetic coupling between both coils taking in account the minimum required distance of both coils.

## IV. Conclusion

Nasogastric tube misplacement can have serious, life-threatening consequences, including death. This work examines a novel device, based on magnetic inductive coupling, for nasogastric tube tip detection which has the potential to reduce serious injury and death due to misplacement of nasogastric tubes. The design parameters are investigated based on the limitations imposed by the size of the nasogastric tube and the expected range of distances. Additionally, preliminary studies were performed to optimise the size of the transmitter coil. It was observed that the choice of the transmitter coil radius can influence the detection of the receiver coil. In particular, depending on the expected distances between the two coils in use, the size of the transmitter coil can be optimised to maximise detection, for example, at distances of 60 mm and above, a 100 mm radius transmitter coil maintains the highest coupling although the same coil is less efficient than smaller radius coils at closer distances.

These results suggest a number of areas for further investigation. Firstly, even in ideal conditions, the relative sizes of the coils can impact the detection depth, particularly given the constraints on the receiver coil in the nasogastric tube itself. Secondly, further analysis of the coupling between the coils in a variety of orientations are required to estimate the locations and orientations of detection. Finally, three-dimensional detection approaches need to be considered to account for movement of the nasogastric tube while in position.

## Acknowledgment


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